

PCT

WORLD INTELLECTUAL PROPERTY ORGANIZATION
International Bureau



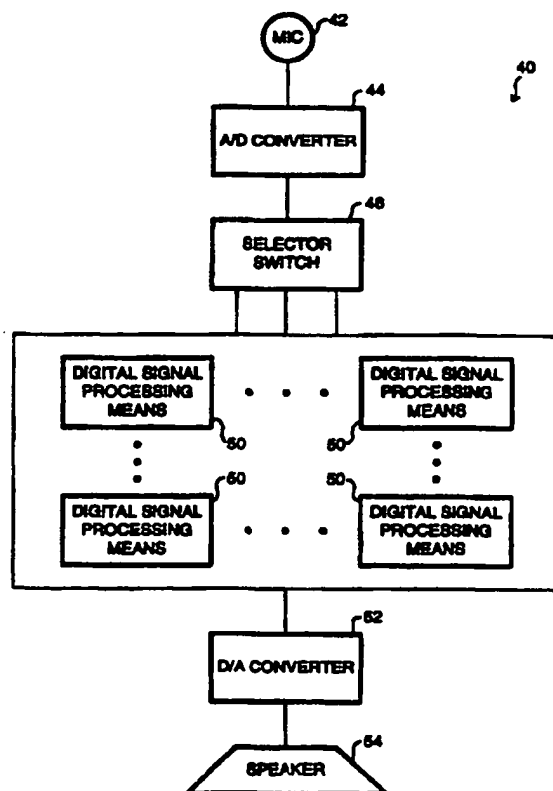
INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(51) International Patent Classification ⁶ : H04R 25/00		A2	(11) International Publication Number: WO 97/14266
			(43) International Publication Date: 17 April 1997 (17.04.97)
(21) International Application Number: PCT/US96/15423			(81) Designated States: AL, AM, AT, AU, AZ, BB, BG, BR, BY, CA, CH, CN, CU, CZ, DE, DK, EE, ES, FI, GB, GE, HU, IL, IS, JP, KE, KG, KP, KR, KZ, LK, LR, LS, LT, LU, LV, MD, MG, MK, MN, MW, MX, NO, NZ, PL, PT, RO, RU, SD, SE, SG, SI, SK, TJ, TM, TR, TT, UA, UG, UZ, VN, ARIPO patent (KE, LS, MW, SD, SZ, UG), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, ML, MR, NE, SN, TD, TG).
(22) International Filing Date: 26 September 1996 (26.09.96)			
(30) Priority Data: 08/540,534 10 October 1995 (10.10.95) US			
(71) Applicant: AUDIOLOGIC, INC. [US/US]; Suite 200, 6655 Lookout Road, Boulder, CO 80301 (US).			
(72) Inventors: MELANSON, John, L.; 7973 Sagebrush Court, Boulder, CO 80301 (US). LINDEMANN, Eric; 2975 18th Street, Boulder, CO 80301 (US).			
(74) Agents: McNELIS, John, T. et al.; Fenwick & West L.L.P., Suite 700, Two Palo Alto Square, Palo Alto, CA 94306 (US).			Published <i>Without international search report and to be republished upon receipt of that report.</i>

(54) Title: DIGITAL SIGNAL PROCESSING HEARING AID WITH PROCESSING STRATEGY SELECTION

(57) Abstract

A digital signal processing hearing aid is disclosed having a plurality of digital signal processing means (50) for processing input digital signals, and a selector switch (46) manipulable by a user for choosing which of the processing means to utilize. Each of the digital signal processing means is designed to provide optimal results in a particular listening environment. Since the user is allowed to choose which of the plurality of processing means to invoke, and since each processing means is specifically designed to operate in a particular listening environment, the hearing aid is capable of providing excellent results in a plurality of listening environments.



FOR THE PURPOSES OF INFORMATION ONLY

Codes used to identify States party to the PCT on the front pages of pamphlets publishing international applications under the PCT.

AM	Armenia	GB	United Kingdom	MW	Malawi
AT	Austria	GE	Georgia	MX	Mexico
AU	Australia	GN	Guinea	NE	Niger
BB	Barbados	GR	Greece	NL	Netherlands
BE	Belgium	HU	Hungary	NO	Norway
BF	Burkina Faso	IE	Ireland	NZ	New Zealand
BG	Bulgaria	IT	Italy	PL	Poland
BJ	Benin	JP	Japan	PT	Portugal
BR	Brazil	KE	Kenya	RO	Romania
BY	Belarus	KG	Kyrgyzstan	RU	Russian Federation
CA	Canada	KP	Democratic People's Republic of Korea	SD	Sudan
CF	Central African Republic	KR	Republic of Korea	SE	Sweden
CG	Congo	KZ	Kazakhstan	SG	Singapore
CH	Switzerland	LI	Liechtenstein	SI	Slovenia
CI	Côte d'Ivoire	LK	Sri Lanka	SK	Slovakia
CM	Cameroon	LR	Liberia	SN	Senegal
CN	China	LT	Lithuania	SZ	Swaziland
CS	Czechoslovakia	LU	Luxembourg	TD	Chad
CZ	Czech Republic	LV	Latvia	TG	Togo
DE	Germany	MC	Monaco	TJ	Tajikistan
DK	Denmark	MD	Republic of Moldova	TT	Trinidad and Tobago
EE	Estonia	MG	Madagascar	UA	Ukraine
ES	Spain	ML	Mali	UG	Uganda
FI	Finland	MN	Mongolia	US	United States of America
FR	France	MR	Mauritania	UZ	Uzbekistan
GA	Gabon			VN	Viet Nam

DIGITAL SIGNAL PROCESSING HEARING AID WITH PROCESSING STRATEGY SELECTIONBackground of the Invention

5 A common problem associated with sensorineural hearing loss is recruitment. A hearing impaired person suffering from recruitment has an elevated threshold for soft sounds. This means that soft sounds which are audible to a person with normal hearing will have to be made louder in order to be heard by the hearing impaired person. However, with recruitment, loud sounds may be just as loud for the hearing impaired person as for the person with normal
10 hearing. This represents a loss of dynamic range for the hearing impaired. This loss of dynamic range may vary with frequency. For example, at low frequencies the hearing impaired person may have nearly the same dynamic range as the person with normal hearing, but at high frequencies the dynamic range of the hearing impaired person may be considerably reduced. This impaired dynamic range is often referred to as the residual dynamic range.

15 The loss of dynamic range in the hearing impaired is most often attributed to malfunction of the outer hair cells of the cochlea. Sound vibrations in the air are transmitted from the ear drum and through the ossicles of the middle ear to the inner ear and the cochlea. Inside the cochlea are the flexible tectorial membrane and the more rigid basilar membrane. Between these two membranes lie the inner and outer hair cells. Ninety-five percent of the afferent neural fibers
20 which transmit acoustic information to the brain are connected to the inner hair cells. The longest cilia of the outer hair cells are connected to the tectorial membrane, but the inner hair cells have no such connection. Both the inner and outer hair cells are connected to the basilar membrane through supporting cells. Vibrations passing between the tectorial and basilar membranes cause more motion in the flexible tectorial membrane than in the basilar membrane.
25 This difference in motion causes a sheering motion along the outer hair cells. The outer hair cells react to this shearing motion in a complex manner. The entire mechanism is not yet clearly understood but it appears that the outer hair cells stretch and contract according to the intensity of the vibrations in a manner which amplifies these vibrations. For larger amplitude vibrations, however, the outer hair cell motion saturates causing a reduction in amplification. This
30 nonlinear, saturating amplification corresponds to a natural dynamic range compression. The compressed vibrations from the outer hair cells are communicated to the inner hair cells and then through the afferant neural fibers to the brain. When the outer hair cells malfunction, there is a loss of natural compression and recruitment occurs. The inner hair cells may continue to functions normally and there may be a mild to moderate hearing loss. More severe hearing
35 losses will occur with loss of inner hair cell function.

 Many hearing aid instruments have been designed to deal with this problem. The approach taken is to compress the dynamic range of the input sound signal so that it more nearly fits into the residual dynamic range of the recruited ear. The ratio of input dynamic range in dB to compressor output dynamic range in dB is called the compression ratio. To adequately specify

the compressor, the compression ratio needs to be accompanied by a static gain value. This static gain value will determine at which input power level the system delivers a specified fixed gain. For example the static gain may be set so that at 80dB SPL input power, the system delivers unity gain. If the compressor is set to a 2:1 compression ratio, then at 60dB SPL input power the system will produce a 70dB SPL output, that is a gain of 10dB, and at 100dB SPL input power the system will produce a 90dB SPL output, that is a gain of -10dB.

Usually the compression ratio is not constant over the entire input power range. A low level compression knee may be defined. For input powers below this low level compression knee, the compression ratio may be 1:1, that is, a fixed linear gain may be applied. The designated compression ratio (e.g. 2:1) may take affect only for input power levels above this low level compression knee. A high level or limiting knee may also be defined. For input power levels above this high level knee, the compression ratio may increase or even become infinite, or it may be that the output level is fixed regardless of increase in input level. A system which has only a high level compression knee below which the compression ratio is 1:1 (linear gain) is called a limiter. A system which has a low level compression knee positioned at 40-50dB SPL is termed a full range compressor.

Even without reference to the electro-mechanics of the inner ear and the natural loss of compression due to malfunction of the outer hair cells, the need for compressors or limiters in hearing aids has long been recognized. The need for hearing aids to have large gains to make softer sounds audible has driven amplifiers and output transducers out of their linear ranges. Earlier hearing aids accomplished limiting by letting the amplifier and/or output transducer clip. Unfortunately this caused harmonic distortion which, at high frequencies, masked softer speech sounds and generally reduced fidelity in the system (See M. C. Killian, *The K-Amp Hearing Aid: An Attempt to Present High Fidelity for Persons with Impaired Hearing*, American Speech-Language-Hearing Association, July 1993, at 52-74). Later systems introduced limiters to help alleviate this problem, and still later systems used full range dynamic range compression (See e.g. Fred Waldhauer et al., *Full Dynamic Range Multiband Compression in a Hearing Aid*, The Hearing Journal, Sept. 1988, at 1-4).

The compression process requires a means for measuring the power of the input signal and generating a dynamically varying gain as a function of this input power. This gain is then applied to the signal which is delivered to the ear. When the input power is low, this gain will generally be high so that soft sounds are made louder. When the input power is high, this gain will generally be low so that loud sounds are not made too loud. The measure of input power requires averaging over time. The time span of the averaging defines a compression time constant. If the time span is very long then the compressor will react slowly to changes in input power level. This is sometimes referred to as Automatic Gain Control (AGC) where time constants of one to two seconds are typical. When the time span of the averaging is short the compressor will react quickly to changes in input power level. With a time span of approximately five to fifty milliseconds, the compressor may be referred to as a syllabic rate

compressor. A syllabic rate compressor will limit the gain of a loud vowel sound while amplifying a soft consonant which immediately follows it.

In most designs there is both an attack and release compressor time constant. The attack time constant determines the time it takes for the compressor to react at the onset of a loud sound. That is, the time it takes to turn down the gain. The release time constant determines the time it takes for the system to turn up the gain again after the loud sound has terminated. Most often the attack time is quite short (<5 milliseconds) with the release time being longer (anywhere from 15 to 100s of milliseconds).

Even with separate attack and release times, there have still been problems with compressor time constants. With a long release time, any short impulse in the room (e.g. the clank of a dish) will cause the gain to be shut down for the length of the relatively long release time. On the other hand, if the time constant is always short, it will cause an annoying swell in volume every time a speaker takes a breath. This problem has been alleviated by the introduction of adaptive time constants. Hotvet introduced in U.S. Patent No. 4,718,499 an adaptive time constant system in which the release time constant for a loud sound in silence is short but the release time constant gradually becomes longer proportional to the length of the louder sounds in the environment. Thus, if a speaker speaks in a normal rhythm, the release time constant will grow longer, reducing the amplitude swell in the brief silences between words. Others have also discussed multiple time constant systems with a similar goal in mind (See e.g. R. F. Laurence, et al., *A Comparison of Behind-the Ear High-Fidelity Linear Hearing Aids and Two-Channel Compression Aids, in the Laboratory and in Everyday Life*, Br. J. Audiol., 1983, at 17:31-48; and Brian Moore, et al., *Optimization of a Slow-Acting Automatic Gain Control System for Use in Hearing Aids*, Br. J. Audiol., 1991, at 25:171-182).

To match the variability of recruitment with frequency, a compressor is often designed to perform differently in different frequency bands. A multi-band compressor divides the input signal into multiple frequency bands and then measures power in each band and compresses each band separately with possibly different compression ratios and time constants in the different bands. For example a properly designed two band compressor can make soft high frequency consonants audible while suppressing low frequency competing noises occurring simultaneously. Vilchur (See E. Vilchur, *Signal Processing to Improve Speech Intelligibility in Perceptive Deafness*, J. Acoust. Soc. Am. 53, 1973, at 1646-1657) discussed a bench top prototype of a two band compressor. Barfod (See J. Barfod, *Multichannel Compression Hearing Aids*, Report No. 11, The Acoustic Laboratory, Technical University of Denmark, 1976) discussed compressors of up to four bands. These compressors also had variable time constants in the different frequency bands.

The outer hair cells of the cochlea, when functioning normally, are often thought to perform compression function in overlapping frequency bands called critical bands. These frequency bands are spaced linearly at intervals of approximately 100 Hz at frequencies below about 500 Hz, and are spaced logarithmically at approximately third octave intervals above

500Hz. Thus, the outer hair cells behave as a biological critical band compressor. The time constant associated with this compressor has been approximated to be about 1ms. Lippman et. al. (See R. P. Lippman, et al., *Study of Multichannel Amplitude Compression and Linear Amplification for Persons with Sensorineural Hearing Loss*, J. Acoust. Soc. Am. 69(2), Feb. 1981, at 524-534) designed a benchtop 16 band compressing hearing aid system with the bands tuned to match the critical bands of hearing. Each band represented a separate compression channel. Two settings of this compressor were compared against a linear non-compressing system. Martin (See G. R. Martin, *Studies of Real-Time Multiband Adaptive Gain Hearing Aids*, MIT, Sept. 1992, at 1-103) discussed a 3rd octave band compression hearing aid system using digital signal processing.

As the number of compression bands increases, each with its own compression ratio and static gain, it is possible to view the compressor as having an almost continuously varying compression ratio as a function of frequency. In this case the system may be represented as a set of frequency dependent gain curves. Each gain curve applies at a certain input power level. For input between these power levels, the system interpolates between gain curves. Killian (previously cited) discusses the K-amp hearing aid system which integrates power in one band but uses the power estimate to interpolate between low level and high level frequency response curves. The low power level frequency response curve has generally more gain and, in particular, more gain at high frequencies than at low. The high power level frequency response curve has generally less gain and is more flat across frequencies. There is an optional setting which allows the low power level curve to also be set flat.

The process of adjusting the compression ratios or gain curves of a compressor is central to the hearing aid fitting process. One approach to doing this is to attempt to adjust the compressor so that for all input levels and all frequencies the hearing impaired listener has the same impression of loudness that a normal listener would have. Loudness is a perceptual quantity which can under certain constraints be plotted as a function of input power level. The loudness growth curve may be measured by presenting a number of input signals at different levels and asking the listener to subjectively rate these on a perceptual scale (e.g. 1 to 10). By measuring the loudness growth curves of an impaired listener at different frequencies and comparing these to the loudness growth curves of an average of normal listeners, a loudness matching compression fitting can be attempted. To accurately match loudness growth curves, the hearing instrument would permit continuously variable compression ratio over input level. In this case it is more useful to think in terms of continuously variable input/output power curves. The system described above with low and high level compression knees is able to implement only three segment piecewise input output curves. Barfod (previously cited) and Lippman et. al. (previously cited) attempted to fit their multi-band compression systems so as to restore the loudness growth curves of the impaired ear to match those of the normal ear.

Loudness matching compression fitting has its limits. If the recruited ear has 5dB of residual dynamic range it will not be effective to compress a 90 dB input dynamic range into this

5dB. Instead, some amount of compression will be applied and then a static gain defined so that the most useful part of the input dynamic range (e.g. typical speech range) is roughly centered in the residual dynamic range. Limiting will be applied for louder signals. Finding good compromises in fitting compressors is central to the art of hearing aid fitting.

5 There has been some discussion about whether it is indeed necessary to test the loudness growth curves of the impaired listener as part of the fitting process or whether it is possible to predict them from the threshold audiograms. Kollmeier et al. (See B. Kollmeier, et al., *Speech Enhancement by Filtering in the Loudness Domain*, Acta Otolaryngol (Stockh) 1990, Suppl. 469:207-214) has shown that the shape of loudness growth curves becomes less predictable with
10 increasing hearing loss. That is, the variance between subjects increases with hearing loss. This indicates that successful prediction from the threshold is unlikely.

 There has been much discussion regarding the nature of improvements due to compression. Vilchur (previously cited) and Yanick (See P. Yanick, *Effects of Signal Processing on the Intelligibility of Speech in Noise for Subjects Possessing Sensorineural Hearing Loss*, J. Am. Audiol. Soc. 1, 1976, at 229-238) showed improvements in intelligibility with their
15 compression systems, while Abramovits (See R. Abramovits, *The Effects of Multichannel Compression Amplification and Frequency Shaping on Speech Intelligibility for Hearing Impaired Subjects*, Unpublished doctoral thesis, City University of New York, 1979), Mangold et al. (See S. Mangold, et al., *Programmable Hearing Aid with Multichannel Compression*, Scand.
20 Audiol. 8, 1979, at 121-126), O'Loughlin (See B. O'Loughlin, *Evaluation of a Three Channel Compression Amplification System on Hearing-Impaired children*, Aust. J. Audiol. 2, 1980, at 1-9), and Lippman et al. (previously cited) failed to show intelligibility improvements. It has also been argued in Moore (See Brian Moore, *Evaluation of a Dual-Channel Full Dynamic Range Compression System for People with Sensorineural Hearing Loss*, Ear and Hearing, Vol. 13, No.
25 5, 1992, at 349-370) that it is necessary to evaluate improvement by testing in the real world for sustained periods of time. Plomp (See Reinier Plomp, *The Negative Effect of Amplitude Compression in Multichannel Hearing Aids in the Light of the Modulation-Transfer Function*, J. Acoust. Soc. Am. 83(6), June 1988, at 2322-2327) has suggested that multi-band compression would be detrimental to speech intelligibility because the reduction in dynamic range does not
30 imply a reduction in the size of the just noticeable difference (JND) in amplitude discrimination. Plomp has further suggested that fast time constant compression would lead to reduced amplitude modulation over time, which in turn, would lead to reduced perception of this modulation. It has also been suggested that very fast time constants can create harmonic distortion at low frequencies. The argument was also put forward that fast time constant multi-
35 band compression would reduce spectral contrasts over frequency, thus "whitening" the spectrum, thereby lessening the ability to distinguish vowels. Vilchur (See E. Vilchur, *Comments on the Negative Effect of Amplitude Compression in Multichannel Hearing Aids in the Light of the Modulation-Transfer Function*, J. Acoust. Soc. Am. 86(1), July 1989, at 425-428) responded to these points. Others have written on related topics. See e.g. L. D. Braida, et al.,

Review of Recent Research on Multiband Amplitude Compression for the Hearing Impaired, The Vanderbilt Hearing Report, Upper Darby, PA: Monographs in Contemporary Audiology, 1982, at 133-140; B. R. Glasberg, et al., *Auditory Filter Shapes in Subjects with Unilateral and Bilateral Cochlear Impairments*, J. Acoust. Soc. Am. 79, 1986, at 1020-1033; Brian Moore, *How Much Do We Gain by Gain Control in Hearing Aids?*, Acta Otolaryngol, 1990, 469 Suppl. at 250-256; Igor Nabelek, *Performance of Hearing-Impaired Listeners under Various Types of Amplitude Compression*, J. Acoust. Soc. Am. 74(3), Sept. 1983, at 776-791; and Walker et al., *The Effects of Multichannel Compression/Expansion Amplification on the Intelligibility of Nonsense Syllables in Noise*, J. Acoust. Soc. Am. 76(3), Sept. 1984, at 746-757.

10 Most agree that some form of limiting is required so that loud sounds are not too loud but soft sounds are audible. The debate is focused on full range vs. limiting compression, and on fast vs. slow time constants. Moore (previously cited) suggests that a two or three band compressor, while having sufficient frequency resolution to allow attenuation of low frequency noise and vowel sounds, and while permitting amplification of softer high frequency consonants, is still
15 coarse enough in frequency, as opposed to a critical band compressor, such that spectral whitening will not occur.

Given two input signals of equal energy, one narrow band so that its frequency range is entirely within one critical band, and another wide band so that its frequency range spans several critical bands, the wide band signal will appear louder to the listener. This is due to a
20 psychoacoustic phenomenon called loudness summation. This has implications for compressor design. If the compressor has a few wide bands (e.g. 2), and if the compressor is adjusted such that wide band signals are well matched in loudness to normal loudness growth curves, then narrow band signals will appear too soft. Conversely if the compressor has many independent narrow bands (e.g. critical bands), and if the compressor is adjusted such that narrow band
25 signals are well matched in loudness to normal loudness growth curves, then wide band signals will appear too loud. Hohman (See V. Hohman, *Narrow/Wide Band Compensation in Coupled Narrow Band Aid*, Reihe 17: Biotechnik, Nr. 93, 1993, at 1-99) has designed a compressor which addresses this problem. It measures not only power but bandwidth of input signals and adjusts gain accordingly. This is called a coupled narrow band compressor.

30 As illustrated by the above discussion, different signal processing strategies have been developed to address different and specific hearing aid problems. In an attempt to increase the versatility of hearing aids, adjustable hearing aids have been developed. With adjustable hearing aids (which typically employ analog signal processors), certain parameters can be adjusted by the user. By allowing the user to dynamically set the parameters, the adjustable hearing aid allows
35 the user to set the hearing aid to best suit the user's listening environment. While an adjustable hearing aid does impart to the user a greater degree of versatility, this versatility has its limits. Ultimately, an adjustable hearing aid implements the same signal processing strategy, regardless of the parameters. If the particular strategy implemented by the hearing aid is not well-suited for

a particular situation, then no amount of parameter adjustment will cause the hearing aid to provide satisfactory results.

Summary of the Invention

5 The present invention is based, at least partially, on the observation that, given all of the available signal processing strategies and all of the possible listening environments that a user may find himself or herself in, there is no single signal processing strategy which provides optimal performance in all situations. In order to be optimal, a hearing aid needs to implement different signal processing strategies for different situations, with each strategy designed for a particular situation. The present invention provides just such a hearing aid.

10 In accordance with the present invention, there is provided a hearing aid comprising an input transducer, an analog-to-digital converter, a plurality of digital signal processing means, a processing means selector manipulable by a user, a digital-to-analog converter, and an output transducer. Preferably, each of the digital signal processing means implements a particular processing strategy designed for a particular situation. The processing means selector allows the user to select which digital signal processing means to invoke so that the user may dynamically choose the best processing means, and hence, the best strategy for any particular listening environment. In a preferred embodiment, the plurality of digital signal processing means of the hearing aid of the present invention is implemented by way of a logic unit and a multi-program store for storing a plurality of instruction sequences. These instruction sequences, when executed by the logic unit, cause the logic unit to implement the functions of the various digital signal processing means. To change digital signal processing means, all that needs to be done is to have the logic unit execute a different set of instruction sequence. Hence, the present invention provides a hearing aid capable of: (1) implementing a number of different signal processing strategies; and (2) allowing a user to select which strategy is implemented, thereby allowing the user to choose the best strategy for any given situation. Overall, the present invention provides a functionally superior hearing aid.

Brief Description of the Drawings

30 Figure 1a is a block diagram representation of the hearing aid of the present invention.

 Figure 1b is a block diagram of a preferred embodiment of the hearing aid of Figure 1a.

 Figure 2 is a functional block diagram of one of the digital signal processing means 50 of Figure 1a.

35 Figure 3 is a functional diagram of an incremental FFT filter bank in accordance with the present invention.

 Figure 4 is a functional diagram of a hybrid filter bank in accordance with the present invention.

 Figure 5 is a plot of the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 306 of Figure 3.

Figure 6 is a plot of the composite frequency response seen at the output of adder 306 of Figure 3.

Figure 7 is a plot of the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 307 of Figure 3.

5 Figure 8 is a plot of the composite frequency response seen at the output of adder 307 of Figure 3.

Figure 9 is a plot of the superimposed magnitude frequency responses of comb filters 301, 303 and 305, 308 of Figure 3.

10 Figure 10 is a plot of the composite frequency response seen at the output of adder 308 of Figure 3.

Figure 11 is a plot of the three superimposed frequency responses shown in Figures 5, 7, and 9.

Figure 12 is a composite frequency response of port out2 shown in Figure 3.

15 Figure 13 is a plot of the magnitude frequency response of a complex filter tuned to FSAMP/4 superimposed on the frequency response seen at the output of adder 406 shown in Figure 4.

Figure 14 is a plot of the composite frequency response seen at the output of the complex one-pole 441 shown in Figure 4.

20 Figure 15 is a plot of the group delay of a complex one-pole resonator with a .9 coefficient in a hybrid filter bank equivalent to a 256 point FFT.

Figure 16 is a block diagram representation of a bandsplitter in accordance with the present invention.

Figure 17 is a block diagram representation of an allpass bandmerger in accordance with the present invention.

25 Figure 18 is a functional representation of an arithmetic logic unit in accordance with the present invention.

Figure 19 is a functional representation of a first embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein a midband notch/bandpass filter is utilized.

30 Figure 20 is a functional representation of a second embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein a midband notch/bandpass filter is utilized, and wherein a real bandsplit is performed before converting to complex.

Figure 21 is a functional representation of a third embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein no midband notch/bandpass filter is utilized, and wherein a real bandsplit is performed before converting to complex.

35 Figure 22 is a functional representation of one of the one-band compressors of a multi-band compressor in accordance with the present invention.

Figure 23 is a more detailed functional diagram of the log smoother 2203 shown in Figure 22.

Figure 24 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 19.

Figure 25 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 20.

5 Figure 26 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 21.

Figures 27-32 are spectra patterns of various signals showing the results of decimation.

Figure 33 is a functional representation of a Hilbert transformer in accordance with the present invention.

10 Figure 34 is a functional representation of a system which results when the bandsplitter shown in Figure 16 is provided with a complex input and is partitioned into its real and imaginary parts.

Detailed Description of the Preferred Embodiments

15 With reference to Figure 1a, there is shown a block diagram representation of the hearing aid of the present invention, the hearing aid 40 preferably comprising an input transducer 42 (preferably taking the form of a microphone), an analog-to-digital converter 44, a selector switch 46, a plurality of digital signal processing means 50, each selectable by selector switch 46, a digital-to-analog converter 52, and an output transducer 54 (preferably taking the form of a speaker). The selector switch 46 is preferably manipulable by a user to allow the user to
20 dynamically select which of the digital signal processing means 50 to invoke in which listening environment. Preferably, each of the digital signal processing means 50 is specifically and optimally designed to deal with a particular listening environment. For example, one of the digital signal processing means 50 may be designed to compensate for noisy environments, while
25 another may be tailored for quiet environments. In dealing with these environments, each of the processing means 50 may implement such functions as compression, noise compression, feedback cancellation, etc.

The hearing aid of Figure 1a operates by receiving audio signals, via the microphone 42, from a particular listening environment, and converting these audio signals into a set of analog
30 electrical signals. These electrical signals are then converted by the analog-to-digital converter 44 into an input digital signal or stream. The input digital stream is then fed to one of the digital signal processing means 50 selected by the selector switch 46, where the input digital stream is processed to derive an output digital signal or stream. This output digital stream is thereafter processed by the digital-to-analog converter 52 to derive a set of output analog electrical signals.
35 Once derived, the analog electrical signals are used to drive the speaker 54 to cause the speaker to produce a set of audio signals which can be heard by the hearing aid user.

A significant advantage provided by the hearing aid of the present invention is that it is capable of implementing a plurality of different rehabilitation strategies. This is in sharp contrast to the adjustable hearing aids of the prior art which are capable of implementing only a single

strategy with adjustable parameters. Because the hearing aid of the present invention can implement more than one strategy (i.e. can change from one strategy to another), it is better able to adapt and to provide optimal results in a variety of different listening environments. As used herein, the expression "changing strategies" means generally switching from one digital signal processing means 50 to another. In actuality, what is often changed in going from one processing means to another is the number of bandpass signals into which the input digital signal is divided, and the bandwidths of these bandpass signals. Thus, by changing strategies, the hearing aid of the present invention is in effect changing the specifications of the filter banks of the digital signal processing means 50. This will become more clear as the invention is described in greater detail.

Referring now to Figure 1b, there is shown a preferred embodiment of the hearing aid of the present invention. Figure 1b shows the main components of the hearing aid 40. This architecture is suitable for implementing a number of dynamic range compression strategies as well as other hearing aid rehabilitation strategies. Sound is input through an input transducer (101) and converted to a digital input stream by the Analog to Digital Converter (102). Calculations are performed on data from the input stream as well as data stored in X Data Ram (105) and Y Data Ram (106). These calculation are carried out by the Arithmetic Logic Unit (108) with the Input Mux (107) selecting which sources of data will be processed. The results of the calculations are fed back to the X or Y Data Rams (105, 106) or to the Digital to Analog Converter (104) which converts the signal to an analog output electrical signal suitable for driving the Output Transducer (103). Corresponding to each hearing aid rehabilitation strategy is a digital signal processing program which is stored in the Multi-program Store (111). Each program corresponds to a set of instructions which are interpreted by the program sequencer (110) and executed by ALU 108 to cause various actions to take place within the rest of the circuit. The Program Selection Switch (112) selects which rehabilitation strategy to activate. This switch is under control of the hearing aid wearer so that as he or she enters different listening environments, the appropriate strategy can be selected. At this point, it should be noted that each of the digital signal processing means 50 shown in Fig. 1a is implemented in the preferred embodiment as a digital signal processing program executed by the ALU 108. By switching between processing programs using program selection switch 112, the hearing aid wearer is in effect switching between the plurality of digital signal processing means 50.

A number of rehabilitation strategies are implemented as digital signal processing programs stored in the Multi-program Store. Figure 2 shows the basic elements common to all of the rehabilitation strategies. The sound signal is input to a Filter Bank Analyzer (201) which divides the signal input into multiple frequency bands. The multiple frequency band signals are input to the multi-band processor (202) which processes the individual frequency band signals to affect their dynamic ranges. In the preferred embodiment, processor (202) performs a compression function; however, it should be noted that processor (202) may perform any desired function. The processed frequency band signals are then input to the Filter Bank

Synthesizer/Combiner (203) which recombines the individual frequency band signals into a single output. Since these basic elements are implemented as digital signal processing programs, and the system provides for a plurality of programs that the user can switch between, it is possible to change filter structures by selecting different programs. For example, some
 5 algorithms, such as noise suppression algorithms, may require fine frequency resolution filter banks, whereas more simple compressors may require only two bands. These differences can be accommodated by switching digital signal processing programs. The following sections describe embodiments related to these three basic components. One of the motivations for the various filter bank embodiments is to achieve high frequency resolution, especially at low frequencies,
 10 without incurring large delay through the system. This issue will be discussed in conjunction with the various embodiments.

INCREMENTAL FFT FILTER BANK ANALYZER

Figure 3 describes the first embodiment of a filter bank analyzer. The sum of a signal and
 15 a delayed version of itself form a comb filter with N filter lobes evenly spaced across frequency from 0 to the sample frequency (FSAMP). The peaks of the lobes of the comb filter are centered at $k \cdot \text{FSAMP}/N$, with k ranging from 0 to N-1. The difference of a signal and a delayed version of itself also forms a comb filter with N evenly spaced lobes but now the peaks of the lobes are centered at $(k \cdot \text{FSAMP}/N) + \text{FSAMP}/(N \cdot 2)$, so that the lobe centers are shifted in frequency by
 20 half a lobe width. The magnitude frequency response of the sum of these two comb filters is flat, that is the comb filters are complementary and together they define $2 \cdot N$ frequency lobes ranging from 0 to FSAMP. A more general form of the comb filter is:

$$S + \text{delay}(S, N) \cdot \text{cnn} \quad (1)$$

where S is the input signal, $\text{delay}()$ is a function which delays the signal by input parameter N,
 25 and cnn is a multiplier coefficient defined as :

$$\text{cnn} = e^{(j \cdot 2\pi \cdot N \cdot \text{CENTER_FREQUENCY} / \text{FSAMP})} \quad (2)$$

which shifts the peak of the first frequency lobe to CENTER_FREQUENCY.

In Figure 3 the numbers c40, c41, c20, c21, c10, c11 etc. represent multiplier coefficients as defined in (2). Thus, the output of adder 302 is a comb filtered signal with delay N defined by
 30 301 as 4. In this case $c40=1$ and $c41=-1$ so the output of adder 303 is the complementary comb filter with lobe centers shifted by $\text{FSAMP}/(4 \cdot 2)$. The output of 302 is then fed into another pair of complementary comb filters whose outputs are the adders 306 and 307 and multipliers $c20=1$ and $c21=-1$. The second stage comb filters defined by 304, 306, 307 and multipliers $c20, c21$ have lobe widths which are twice the width of the first stage comb filter with output 302 since
 35 304 has delay 2 which is one half delay 301. Since the lobe width of the second stage comb filter 304, 306 is twice the first stage comb filter 301, 302, and since they both have their first lobes centered at frequency 0 then the second stage comb filter will have a zero at the peak frequency of the first stage comb filter's second lobe and all subsequent even number lobes. Figure 5 shows the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 306.

With these two comb filters in series the composite frequency response seen at the output of adder 306 is shown in Figure 6. In effect the second stage comb filter selects lobes 1, 3 of the first stage comb filter and suppresses lobes 2, 4. The second stage comb 304, 307 is the complement of comb 304, 306 since $c21=-1$ and its lobes are shifted in frequency by half the second stage lobe width, that is by one of the first stage comb widths. Therefore, second stage comb 304, 307 selects lobes 2, 4 of the first stage comb and suppresses lobes 1, 3. Figure 7 shows the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 307 and Figure 8 shows the composite frequency response seen at the output of adder 307.

The frequency response at the output of adder 303 is the complement of that at the output of adder 302 and is shifted by one half the first stage lobe width compared to the output of 302. The outputs of adders 308 and 309 are again complementary comb filters with lobe widths twice the width of the first stage comb, but now to select the even and odd lobes respectively of the output of adder 303 they must be shifted by .5 and 1.5 of the first lobe widths so that they will line up correctly with the output of adder 303. This means that $c22=j$ and $c23=-j$ where $j=\sqrt{-1}$. Thus the outputs of 308 and 309 are complex signals as will be the general case with this network. Figure 9 shows the superimposed frequency responses of combs 301, 303 and 305, 308 and Figure 10 shows the composite frequency response seen at the output of adder 308. At the output of adders 306 through 309 there are 4 frequency responses each one selecting two of the original ($N*2=8$) lobes defined by the first stage comb filters. By continuing this process of doubling the lobe width and shifting lobe centers, a third stage of comb filters with delays of 1 is added providing eight outputs, each selecting one of the original eight lobes as defined by the first stages comb filters. This requires complex multipliers $c10-c17$ selected appropriately. Figure 11 shows the three superimposed frequency responses and Figure 12 shows the composite frequency response leading to out2 which is the third lobe of the original eight.

The outputs of the system of comb filters defined in Figure 3 are identical to those of an 8 point Complex Fourier Transform. In effect the system is the implementation of an Incremental Complex Fourier Transformer with rectangular window. It is incremental because for every new input sample a new set of output transform samples is generated. By adding additional comb filter stages in front of the first stage shown in Figure 3, and by expanding the comb filter tree appropriately, longer Fourier Transforms can be calculated. The number of frequency points of the Fourier Transform is $2*N$ where N is the delay of the first stage comb.

The group delay of the Incremental Complex Fourier Transformer is the sum of the series interconnection of the combs. Each comb filter has a group delay equal to $\frac{1}{2}$ the delay length. So for an $N*2$ point Fourier Transform the total group delay is $N/2 + N/4 \dots + \frac{1}{2}$ which is equal to $N-.5$. Thus, the 8 point FFT system has a delay of $4-.5 = 3.5$ samples. A typical block based implementation of a Short Time Fourier Transform system with a 2 to 1 overlap of successive FFT frames has a total delay of $3*(N)$ where $N*2$ is the FFT window length. This delay is due to system buffering requirements. This is more than 3 times the length of the optimal Incremental

Fourier Transform system. A system for implementing a 256 point FFT would have delays of 384 samples in the block case and 127.5 samples in the Incremental Fourier Transform case.

HYBRID COMB FILTER/RESONATOR FILTER BANK ANALYZER

5 In Figure 4 elements 401 through 409 implement two comb filter stages similar to those described for Figure 3; however, now the delay is $N=8$ so the system should have $2*N=16$ frequency output points.

The frequency responses seen at the output of adders 406 through 409 each select 4 out of the 16 lobes defined by the first stage comb filters. Now, however, instead of continuing the system of comb filters another 2 stages to have sixteen outputs each with 1 lobe selected we
10 instead apply recursive resonator filters to the outputs of adders 406 through 409. In the preferred embodiment, these filters take the form of one-pole complex resonators; however, it should be noted that other types of filters may also be used. The complex one pole filter is defined as:

$$15 \quad y(n) = (1 - c) \cdot x(n) + c \cdot y(n - 1)$$

where all quantities are complex. Figure 13 shows the magnitude frequency response of a complex filter tuned to $FSAMP/4$ superimposed on the frequency response seen at the output of adder 406. Figure 14 shows the composite frequency response of the two filters seen at the output of complex one pole 411. The filter feedback coefficient of the complex one pole was set
20 to .9 in Figure 14. In general the value of the coefficient defines the sharpness of the selected lobe, with sharper lobes giving greater isolation from band to band but more ripple in the total filter bank response. Note that while it is possible to apply 4 different complex one pole filters to the output of each adder 406-409, in fact, only two are applied because it is only desired to acquire samples for frequencies 0 through $FSAMP/2$ of a real input signal. In some situations
25 where the system is applied to a complex input signal, all complex one pole signals can be applied.

In general, given an Incremental Fourier Transform system of arbitrary order, it is possible to cut the system after some number of comb filter stages and then apply complex one pole resonators to select frequency lobes. A system for implementing the equivalent of a 256
30 point FFT, that is 128 frequency bands from 0 to $FSAMP/2$, can have two stages of comb filters followed by 128 complex resonators. The group delay is equal to the series connection of the two comb filter stages followed by the complex resonator. Figure 15 shows the group delay of the complex one pole with .9 coef. It has a peak value of 9 samples and an average value of approximately 1 sample. The overall delay of the 256 point system would be $128/2 + 64/2 + 9 =$
35 104 in the worst case compared to 127.5 for the Incremental Fourier Transform case. The disadvantage is that the group delay is not flat which can create a reverberation artifact if the feedback coefficient of the complex one poles is too close to one causing very sharp resonators with large peaks in group delay.

Often it is desirable to have filter banks with unequal spacing of bands across frequency. In particular, a spacing similar to the ears' critical bands (approximately 100 Hz spacing below 500 Hz and approximately third octave above) is often desirable. While this kind of quasi-constant Q spacing is not easily achievable with the comb filter structure, it is possible to grossly approximate nonlinear spacing by having more than one comb filter system in parallel. Note that in the Hybrid Comb Filter/Resonator structure with only two comb filter stages, the bulk of the computation is in the complex one pole filters. Therefore, in another embodiment of the Hybrid Comb Filter/Resonator there are two 2 stage comb filter systems in parallel with $N=8$ and $N=4$ respectively for the two systems. The $N=8$ comb filter system has complex one pole filters only in the lower half band, while the $N=4$ comb filter system has complex one pole filters only in the upper half band. It should be clear to one skilled in the art that it is possible to have more than two comb filter systems in parallel, and that the delay lengths for two stage systems can be any even number allowing for a wide variety of nonlinear frequency spacing configurations.

15 HYBRID BANDSPLITTER FILTER BANK ANALYZER

While the Incremental Fourier Transform and Hybrid Comb Filter/Resonator Filter banks are effective for reducing delay compared to a block oriented Short Time Fourier Transform, they are still costly in terms of computation. An approach to alleviating this is to first divide the spectrum into a relatively small number of bands (e.g. 4) and then apply the above mentioned filter bank techniques within those sub-bands that require more frequency resolution, such as the lowest frequency bands. Efficient bandsplitter filters based on allpass filter structures have been proposed (See P. P. Vaidyanathan, MULTIRATE SYSTEMS AND FILTER BANKS, PTR Prentice Hall, Englewood Cliffs, NJ, 1993). The text of this reference is incorporated herein by this reference. The bandsplitter divides a signal into a highpass and lowpass signal with crossover at the mid frequency point of the band. The bandsplitters proposed are power symmetric meaning that the magnitude frequency response of the sum of the two filters is perfectly flat. However the phase response of the filters is not linear so that, like the complex one pole resonators described above, the system will have peaks in the group delay at the crossover bands of the bandsplitters.

Figure 16 shows the structure of the allpass bandsplitter. Since the low and high pass signals each have half the bandwidth of the input signal they can be decimated by a factor of 2 in sample rate. The allpass filters (1604, 1605) for the bandsplitters have the special property that the filter coefficients for odd numbered powers of Z , (Z^{-1}, Z^{-3}, \dots), are all zero. Because of this special property it is possible to move the decimators (1602, 1603) in front of the allpass sections 1604, 1605. In this way the computation rate of the allpass section is halved. In practice, rather than providing explicit decimators, one path is fed with the even number points and the other, because of the unit sample delay operator (1601), receives the odd points. In fact, some aliasing may occur due to this decimation process because the filters are not ideal bandsplitters and have a finite transition band over which aliasing can occur. A discussion of aliasing minimization will

occur later in this section. It has been shown by Vaidyanathan (previously cited) that a ninth order elliptical bandsplitter can be implemented with 2 second order allpass sections.

The intention is to split the original signal into low and high bands and then split the low band again to create low low (LL) bands and low high (LH) bands. Then the LL and LH bands will be further divided by inputting them to separate filter banks either of the Incremental Fourier Transform type as in one embodiment or the Hybrid Comb Filter/Resonator type as in another embodiment.

As described above the two filter bank types operate on complex values; therefore, it can be desirable to convert the real signals to complex signals before inputting them to the filter banks. The conversion to complex also has advantages in terms of quantization of filter coefficients in the allpass bandsplitters.

In any embodiment involving complex signals it is necessary to convert from real to complex. This is done using a Hilbert transformer which generates a pair of signals which are in quadrature (90 degree phase lag) relationship across the usable frequency band. One of the pair is the real part of the complex signal, and the other of the pair, lagging by 90 degrees, is the imaginary part. Together this complex pair is referred to as the analytic signal. It should be noted that referring to this pair as complex is a mere convenience. It is quite possible to generate two real signals in quadrature relationship and then continue to process them in a manner identical to that described in this patent without ever referring to them as complex with an identical result and system structure. Therefore, without loss of generality, this disclosure will continue to refer to complex signals, and it is to be understood by those skilled in the art that this can apply to systems involving real signals in quadrature relationship without a fundamental change of structure.

The spectrum of a real signal is conjugate symmetric with the interval from frequency 0 to $-FSAMP/2$ being the conjugate mirror of the interval from 0 to $FSAMP/2$. An analytic signal generated directly from this signal is ideally identical in the interval 0 to $FSAMP/2$ but the interval 0 to $-FSAMP/2$ is zero. So the ideal Hilbert transformer is a rectangular filter with unity gain from 0 to $FSAMP/2$ and 0 gain from 0 to $-FSAMP/2$. This rectangular shape is the same as an ideal real-valued lowpass filter which has been shifted in frequency by $FSAMP/4$. The allpass bandsplitters described above provide an efficient structure for implementing a lowpass filter. Shifting the frequency response by $FSAMP/4$ should provide the desired Hilbert Transformer. Note that since the Hilbert Transform process results in zeroing half the spectrum, it is then possible in the ideal case to decimate the signal by 2 without loss of information, in which case the interval from 0 to $FSAMP$ after Hilbert Transformation and decimation will be the same as the interval from 0 to $FSAMP/2$ of the original real signal, and this 0 to $FSAMP$ will be repeated at every multiple of the decimated $FSAMP$. In practice, some aliasing will occur due to decimation just as in the bandsplitter case. As with the real bandsplitter described above because of the special properties of the allpass sections used in the Hilbert Transformer it is possible to move the decimators in front of the allpass sections. Figure 33 shows the structure of the Hilbert

Transformer which is seen to be similar to the bandsplitter structure. The shift in frequency is accomplished by multiplying the delay operator 3301 by j and modulating the filter coefficients in 3304 and 3305 by the sequence $\exp(j\pi/2 \cdot n)$, where n is the index of the coefficients beginning with zero. In this sequence, the j terms land on the odd numbered powers of Z which
 5 as we have already indicated are zero for the allpass bandsplitter filters if the decimation occurs after the filters. In this case since the decimation occurs before the filters, the odd numbered zero coefficients simply disappear and the number of coefficients is halved. Therefore, the modulation of the lowpass, highpass real valued filter coefficients is accomplished by negating every other coefficient of the allpass filters when they occur after decimation by 2 as in Figure 33
 10 so that the resulting filters still have all real valued coefficients. Typically the two band outputs of the bandsplitter are formed by taking the sum and difference of the outputs of the two allpass sections as in 1606, 1607 of Figure 16. In the case of Figure 33, since we are interested only in a single Hilbert Transformed complex output, this corresponds to the lowpass, summed output of the frequency shifted allpass sections. Since the input to the upper allpass section (3304) is pure
 15 real, and the input to the lower allpass section (3305) is pure imaginary due to the multiplication by j in 3301, the sum of these two outputs is simply a complex stream made up of the two allpass section outputs: real for the top allpass and imaginary for the bottom allpass. This is the decimated analytic signal.

In several embodiments to be described, the input to a bandsplitter stage is itself complex,
 20 being the output of a Hilbert transformer or a previous complex bandsplitter stage. In this case, the signals are analytic signals which have been decimated by 2 so that the useful pass band of the complex spectrum extends from 0 (lowest frequency) to FSAMP (highest frequency), not FSAMP/2 as for a real signal, and is periodic at intervals of FSAMP as described above. Therefore, splitting the low and high bands requires a lowpass filter with pass band from 0 to
 25 FSAMP/2 and stop band from FSAMP/2 to FSAMP, and a highpass filter with stop band from 0 to FSAMP/2 and pass band from FSAMP/2 to FSAMP. The lowpass filter in this case is seen to be identical to the Hilbert Transformer and the high pass is its complement. If the bandsplitter structure of Figure 16 is provided with a complex input, and if the resulting system is partitioned into its real and imaginary parts, the system of Figure 34 results. Note that since the allpass
 30 coefficients are real-valued, it is possible to process the real and imaginary part of the input separately through two identical pairs of allpass filters 3407, 3408 and 3409, 3410. Taking the sums and differences of the real and imaginary outputs of the allpass sections and pushing the multiply by j of the delay operators (3401, 3402) to the output is equivalent to the summing and differencing network shown in Figure 34 (3411-3414). In particular, no non-trivial complex
 35 multiplies are required to implement the complex bandsplitter and the required computation is exactly twice that of the real bandsplitter with possible additional savings due to identical coefficients in the real and imaginary allpass sections.

The standard DSP Arithmetic Logic Unit (ALU) has two operand inputs to a multiplier and an accumulator following the multiplier. This is an excellent architecture for many DSP

algorithms but the allpass bandsplitters described above are awkward to implement with this structure. However, by placing an additional adder in front of one of the ALU inputs the allpass bandsplitter architecture becomes much simpler to implement. Figure 18 shows the structure of the DSP ALU for the preferred embodiments described herein.

5 Figure 19 shows one embodiment of a Hybrid Bandsplitter Filter Bank Analyzer (HBIFFT). The input signal is first input to the Midband Notch/Bandpass filter (1900) which will notch a small band from the mid frequency point $FSAMP/4$ and deliver this notched band as a separate output from the filter bank. The purpose of this will be described below. The main wideband output of 1900 is fed to the Hilbert Transformer (1901) where it is decimated by 2 and
10 converted to an analytic (complex) signal. This is justified as described above. The output of the Hilbert Transformer (1901) is then fed to the complex bandsplitter (1902) where the high and low bands are generated and each are decimated by 2. The highband complex signal is output from the filter bank and the low band complex signal is fed through another stage of decimation and complex bandsplitting (1903) and then the low low (LL) and low high (LH) bands are fed to
15 separate Filter Banks (1904, 1905). In the embodiment of Figure 19, these filter banks are Hybrid Comb Filter/Resonator types but they can easily be Incremental Fourier Transform types or any other filter bank structure with no loss of generality.

Bandsplitting or Hilbert Transforming a signal and decimating by 2 causes no aliasing in the ideal case. However, in practice, aliasing will occur near the transition bands. Figures 27
20 through 32 show the pattern of spectra for the various cases of decimation. Figure 27 shows the discrete spectrum of a real signal. L indicates the lowest frequency and H indicates the highest frequency at $FSAMP/2$. Figure 28 shows the spectrum after Hilbert transformation and decimation by 2. The $FSAMP$ in Figure 28 is with respect to the decimated sample rate, that is, $FSAMP$ of Figure 28 = $FSAMP/2$ of Figure 27. Note that the highest frequencies of the original
25 spectrum are adjacent to the lowest frequencies and the spectrum extends from 0 to $FSAMP$. In Figure 28, M indicates the mid frequency point halfway between L and H. To the extent that the Hilbert filter is non-ideal and has a non-zero width transition band, there will be aliasing in the overlap region between the highest and the lowest frequencies. In practice, due to limitations of input transducers, the response of any practical hearing aid instrument does not extend down to 0
30 frequency but rather begins at approximately 100 to 150 Hz. Thus, there is a small "don't care" region at the lowest frequencies which can provide a margin of safety from aliasing. Specifically, the low frequencies which fold into the high are nonexistent and the high frequencies which fold into the low can be filtered out provided the folding is below the 100 to 150 Hz band. Since the LL band of lowest frequencies is to be further divided by fine frequency
35 resolution filter bank (1905 in Figure 19) it is possible to zero out the lowest one or two bands of this fine resolution filter bank to accomplish the desired aliasing protection.

The output of the Hilbert Transformer will be fed to a complex bandsplitter. As described above the complex bandsplitter divides the complex spectrum into two half-bands from 0 to $FSAMP/2$ and $FSAMP/2$ to $FSAMP$. The two halfbands are then decimated. Figure

29 and Figure 30 show the lowpass and highpass, respectively, halfband outputs of the complex bandsplitter after decimation. In the low band, L is adjacent to M, the mid frequency point, and in the high band, M is adjacent to H. Since the bandsplitter is non-ideal there will be aliasing around these adjacent regions. The Midband Notch/Bandsplitter (1900 in Figure 19) is responsible for removing a small band region around $M=FSAMP/4$ and sending it as undecimated side information which will be separately processed. This protects against aliasing in any spectra in which the mid frequency point M is adjacent to some other band. Figure 31 and Figure 32 show the low low (LL) and low high (LH) spectra after the lowband has again been bandsplit and each sub-band decimated by 2. In this case M2 indicates the frequency point midway between L and M. Note that M2 is either adjacent to L or to M both of which have aliasing guard regions.

Figure 20 shows a second embodiment of the Hybrid Bandsplitter Filter Bank Analyzer (HBIFFT). Note that the first stage after the Midband Notch/Bandpass (2000) is a real bandsplitter/decimator (2001) followed by subsequent Hilbert Transform/decimator (2002).

Since the high band is not further divided by filter banks involving complex valued comb filters it does not need to be converted to complex. The issues of aliasing avoidance are similar to those for Figure 19.

Figure 21 shows yet a third embodiment of the Hybrid Bandsplitter Filter Bank Analyzer (HBIFFT). Note that the Midband Notch/Bandpass filter has been removed and the first real bandsplitter (2100) has no decimator associated with it so that the real low and high band outputs are undecimated. Having the first bandsplitter outputs be undecimated causes the entire system to be oversampled by a factor of two. Since each half band is undecimated half of the spectrum is zero to within aliasing error. These zero bands travel through the system and protect against aliasing so that a Midband Notch/Bandpass filter is not required.

MULTI-BAND COMPRESSOR

Figure 22 shows the structure of one band of the multi-band compressor. The structure is repeated for every band. An instantaneous power estimate is taken (2200), which for a real input is the square of the input value, and for a complex input is the sum of squares of the real and imaginary parts of the input. The log of the instantaneous power estimate (2201) is then taken. A crude logarithm quantized to the nearest 3db can be taken by simply normalizing the power estimate, that is finding the number of zeros before the first 1 in the double precision power estimate. Any precision can be had by incorporating more bits to the right of the first non-zero bit in the log evaluation process. Extreme precision is not required in this process since the power estimate will be smoothed over time. The most straightforward way to accomplish this smoothing would be to apply a lowpass filter to the linear power estimate before the logarithm is taken. However, since the instantaneous power estimate is double precision due to the squaring operations, the linear smoother would involve double precision multiplications which are costly. To avoid this the smoothing is done after the logarithm is taken.

A general equation which can implement the equivalent of taking the log of the output of a 1 pole recursive linear filter whose input is a linear instantaneous power estimate is:

$$S(n) = f([L(n) - S(n-1)], C) + S(n-1) \quad (3)$$

where:

- 5 $S(n)$ = smoothed log power for sample time n ;
- $L(n)$ = unsmoothed instantaneous log input power;
- C = time constant; and
- $f(x, y)$ = an arbitrary function of 2 inputs;

where $f(x, y)$ is implemented as either a table lookup function or an analytic function of two
 10 inputs. While (3) is able to generate the equivalent of $\log(\text{smooth}(\text{linear power}))$, a close approximation can be attained through:

$$S(n) = f(L(n) - S(n-1)) \cdot C + S(n-1) = \tilde{S}(n) \quad (4)$$

where $\tilde{S}(n)$ is a close approximation to $S(n)$. Figure 23 shows the structure of the Log
 Smoother which implements (4). The difference (2301) of the new instantaneous log estimate
 15 and the current filter state (2306) is fed to a function generator (2302) which in this case is implemented as a lookup table, the output of which is then multiplied (2304) by the smoothing coefficient time constant (2307) and added (2305) to the current filter state (2306) to produce the new smoothed log output which is written back to 2306 as the new filter state.

The Smoothing Coefficient Generator (2307) determines the time constant of the
 20 compressor. Recall that it is generally desirable to have separate attack and release time constants. The Smoothing Coefficient Generator (2307) accomplishes this by comparing the incoming instantaneous power estimate with the current filter state and selecting the appropriate coefficient for attack or release based on this comparison. In addition, it is often desirable to scale the smoothing coefficient depending on the power of the input signal. This is particularly
 25 true when very fine frequency band noise reduction algorithms are implemented. In this case it is desirable to have a longer smoothing time constant for low power signals than for high power signals. The Smoothing Coefficient Generator (2307) also accomplishes this by scaling the gain coefficient as a function of the input power estimate.

In compression algorithms, it is desirable to smooth power estimates in frequency bands
 30 over time. In the related noise reduction patents cited above, this smoothing is critical. In such a case, not only the power estimates, but the cross spectra between the left and right ear signals are also smoothed. This smoothing is accomplished by using a lowpass filter, implemented either as a linear filter or as a logarithmic smoother as described above. If the time constant associated with this lowpass filter is too long, then the signal sounds reverberated. On the other hand, if the
 35 time constant is too short, then there is insufficient smoothing and the signal sounds choppy. This choppiness is most apparent at low input signal power levels, for example, during the silence periods in a conversation where there is an air conditioner or a computer fan in the background. A method for dealing with this problem is to adaptively determine the time constant based on input power level. For low input power, the time constant is made relatively long. For

high input power, the time constant is made relatively short. This serves to prevent the reverberation artifact. The determination of adaptive smoothing time constants may be done in individual frequency bands based on power in the individual bands, or it may be done over the entire passband, with only one smoothing constant being used for all frequency bands.

5

INCREMENTAL FFT FILTER BANK ANALYZER AND HYBRID COMB FILTER/RESONATOR SYNTHESIZERS/COMBINERS

Since the Incremental FFT Filter Bank Analyzer and Hybrid Comb Filter/Resonator Filter Bank Analyzer run at undecimated rates with respect to the sample rate of the input signal to the filter bank, the synthesizers/combiners for both of these analyzers are simple summers which add all the bands of the filter bank to create a single summed output. This adds nothing to the group delay of the filter bank.

10

HYBRID BANDSPLITTER FILTER BANK SYNTHESIZER/COMBINER

The allpass bandsplitters and Hilbert Transformers operate at decimated rates and require synthesizers/combiners to regenerate the output. The synthesizer for the allpass bandsplitter and the Hilbert Transformer is the mirror image of the corresponding analyzer graph. Figure 17 shows the structure of the allpass bandmerger. Like the bandsplitters the interpolate by 2 operation is part of the structure. The structure is the same for real and complex synthesizers but for complex there is a separate identical path for the real and imaginary parts. The bandmerger takes two decimated by 2 inputs and produces one undecimated output. The Inverse Hilbert Transformer is similar except that only one decimated by 2 input is taken. Under ideal undecimated circumstances the Inverse Hilbert Transformer consists simply of taking the real part of the analytic signal. However, because of non-ideal filters leading to aliasing, the Inverse Hilbert Transformer helps to cancel these aliases.

20

Filter 24 shows the structure of the Hybrid Bandsplitter Filter Bank Synthesizer/combiner corresponding to the analyzer shown in Figure 19. It can be seen to be the mirror system with simple summers (2401, 2402) for the high resolution filter banks in the low frequency bands. Filter 25 shows the structure of the Hybrid Bandsplitter Filter Bank Synthesizer/Combiner corresponding to the analyzer shown in Figure 20. It is likewise a mirror system of the analyzer in Figure 20. Figure 26 shows the synthesizer/combiner for the oversampled by 2 case corresponding to the analyzer of Figure 21. In this case the real band merger is a simple summer with no interpolation by 2 since the signals at this point are undecimated.

25

The group delay of the Hybrid Bandsplitter Filter Bank Analyzer and Synthesizer is equal to the sum of all series connected filters. The group delay of the low band filter banks has already been discussed. However, these are now running at $\frac{1}{4}$ or $\frac{1}{8}$ of FSAMP so that the delays in real terms must be multiplied by 4 or 8. In addition the bandsplitters, bandmergers, and Hilbert Transformers and Inverse Transforms all have non constant group delay since they are nonlinear phase. This delay is greatest at the crossover bands of the bandsplitters.

30

PROGRAM SWITCHING

As mentioned previously, the filter banks and compression embodiments disclosed herein are implemented as digital signal processing programs on a programmable digital signal processor. The digital signal processor presents the possibility of completely changing filter bank structures and compression strategies dynamically by loading different digital signal processing programs. Other algorithm types, such as directionality based beamforming noise reduction algorithms may also be loaded. In the current state of the art, the term "programmable hearing aid" refers to a fixed hardware filter and compression structure wherein certain parameters of the fixed structure, such as the compression ratio in each band and the time constants, can be programmed. The hearing aid of the present invention brings new meaning to the term "programmable". The term "programmable" as used herein means "fully software programmable". For a fully software programmable hearing aid, the filtering algorithm is implemented entirely by a program. When a user changes rehabilitation strategies by manipulating the selector switch 112, a different software program is executed by the ALU 108. This new program may entirely change the number of bands, bandwidths, and structure of the filter bank, as well as performing additional functions such as noise suppression. As an example of the desirability of changing filter bank structures, a noise reduction system generally requires many more frequency bands than a compressor. This leads to more power consumption. When noise reduction is not needed, a simpler compression algorithm with a simpler filter bank structure should be used to reduce power consumption. The hearing aid of the present invention allows a user to easily switch from one algorithm to another.

What is claimed is:

1. A hearing aid, comprising:
 - an input transducer for converting audio signals into analog electrical signals;
 - an analog-to-digital converter for converting said analog signals into input digital signals;
 - 5 a plurality of digital signal processing means, each having an input for selectively receiving said input digital signals, and an output for providing a set of output digital signals, each of said digital signal processing means capable of processing said input digital signals to implement a selected filtering strategy designed for a selected situation;
 - a processing means selector manipulable by a user for selecting one of said digital signal
 - 10 processing means to use in processing said input digital signals, said selector enabling the user to dynamically select which of said filtering strategies to implement in any particular situation;
 - a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and
 - an output transducer for converting said output analog electrical signals into a set of
 - 15 output audio signals.
2. The hearing aid of claim 1, wherein at least one of said digital signal processing means comprises:
 - a filter bank analyzer for dividing said input digital signals into a plurality of individual
 - 20 frequency band signals;
 - a multi-band processor for processing said plurality of individual frequency band signals to derive a plurality of processed frequency band signals; and
 - a filter bank combiner for combining said plurality of processed frequency band signals to derive said output digital signals.
 - 25
3. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
 - a first section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and
 - a final section of complementary comb filters for processing said intermediate frequency
 - 30 band signals to provide said plurality of individual frequency band signals.
4. The hearing aid of claim 3, wherein said individual frequency band signals are complex digital signals.
5. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
 - a section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and
 - a section of recursive filter resonators for processing said intermediate frequency band signals to provide said plurality of individual frequency band signals.
 - 35

6. The hearing aid of claim 5, wherein at least one of said recursive filter resonators comprises a complex one-pole recursive digital filter.

5 7. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
a bandsplitter for dividing said input digital signals into two intermediate frequency band signals: a highpass signal and a lowpass signal.

8. The hearing aid of claim 7, further comprising:
10 a filter bank for further dividing at least one of said intermediate frequency band signals into a plurality of further divided frequency band signals.

9. The hearing aid of claim 8, wherein said filter bank comprises:
a plurality of complementary comb filters for processing said one of said intermediate
15 frequency band signals to provide said plurality of further divided frequency band signals.

10. The hearing aid of claim 8, wherein said filter bank comprises:
a section of complementary comb filters for processing said one of said intermediate
frequency band signals to provide a second plurality of intermediate frequency band signals; and
20 a section of recursive filter resonators for processing said second plurality of intermediate frequency band signals to provide said plurality of further divided frequency band signals.

11. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
a midband notch bandpass filter for processing said input digital signals to provide two
25 output signals: a small band signal centered about a frequency midpoint of said input digital signals, and a main wideband signal; and
a bandsplitter for dividing said main wideband signal into two intermediate frequency band signals: a highpass signal and a lowpass signal.

30 12. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
a real bandsplitter for dividing said input digital signals into two intermediate frequency band signals: a real highpass signal and a real lowpass signal; and
a real-to-complex converter for converting said real highpass signal and said real lowpass
signal into complex signals.

35 13. The hearing aid of claim 2, wherein said filter bank analyzer comprises:
a real-to-complex converter for converting said input digital signals into complex digital signals; and

a complex bandsplitter for dividing said complex digital signals into two complex intermediate frequency band signals: a complex highpass signal and a complex lowpass signal.

14. The hearing aid of claim 2, wherein said multi-band processor comprises:
- 5 means for generating a linear instantaneous power estimate for at least one of said individual frequency band signals to produce an instantaneous linear power estimate stream;
- means for converting said instantaneous linear power estimate stream into an instantaneous logarithmic power estimate stream;
- 10 means for smoothing said instantaneous logarithmic power estimate stream to produce a smoothed logarithmic power estimate stream;
- means for calculating a gain coefficient based on said smoothed logarithmic power estimate stream; and
- means for processing said one of said individual frequency band signals and said gain coefficient to generate one of said plurality of processed frequency band signals.

- 15
15. The hearing aid of claim 14, wherein said smoothing means comprises:
- means for processing said instantaneous logarithmic power estimate stream with a smoothing coefficient time constant to derive a processed power estimate stream;
- means for storing a current smoothing filter state;
- 20 means for processing said current smoothing filter state with said processed power estimate stream to derive a new smoothing filter state, said new smoothing filter state being stored in said storing means, and representing said smoothed logarithmic power estimate stream; and
- means for adaptively generating a new smoothing coefficient time constant based on a
- 25 comparison of said current smoothing filter state with said instantaneous power estimate stream.

16. A hearing aid, comprising:
- an input transducer for converting audio signals into analog electrical signals;
- an analog-to-digital converter for converting said analog signals into input digital signals;
- 30 a storage for storing a plurality of instruction sequences, each instruction sequence, when executed, implementing a selected digital signal processing strategy;
- a logic unit, coupled to said analog-to-digital converter, for executing one of said instruction sequences to process said input digital signals in accordance with one of the selected digital signal processing strategies, said logic unit providing a set of output digital signals;
- 35 a program sequencer coupled to said logic unit and said storage for selectively sending one of said instruction sequences to said logic unit for execution thereby;
- an instruction sequence selector, coupled to said program sequencer, manipulable by a user for controlling said program sequencer to select one of said program sequences to send from

said storage to said logic unit, said selector enabling the user to dynamically select which of said digital signal processing strategies to implement in any particular situation;

a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and

5 an output transducer for converting said output analog electrical signals into a set of output audio signals.

17. The hearing aid of claim 16, wherein at least one of said instruction sequences comprises:

10 a filter bank analyzer portion for dividing said input digital signals into a plurality of individual frequency band signals;

a multi-band processor portion for processing said plurality of individual frequency band signals to derive a plurality of processed frequency band signals; and

15 a filter bank combiner portion for combining said plurality of processed frequency band signals to derive said output digital signals.

18. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
a first section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and

20 a final section of complementary comb filters for processing said intermediate frequency band signals to provide said plurality of individual frequency band signals.

19. The hearing aid of claim 18, wherein said individual frequency band signals are complex digital signals.

25

20. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
a section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and

30 a section of recursive filter resonators for processing said intermediate frequency band signals to provide said plurality of individual frequency band signals.

21. The hearing aid of claim 20, wherein at least one of said recursive filter resonators comprises a complex one-pole recursive filter.

35 22. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
a bandsplitter for dividing said input digital signals into two intermediate frequency band signals: a highpass signal and a lowpass signal.

23. The hearing aid of claim 22, further comprising:

a filter bank for further dividing at least one of said intermediate frequency band signals into a plurality of further divided frequency band signals.

24. The hearing aid of claim 23, wherein said filter bank comprises:
5 a plurality of complementary comb filters for processing said one of said intermediate frequency band signals to provide said plurality of further divided frequency band signals.

25. The hearing aid of claim 23, wherein said filter bank comprises:
a section of complementary comb filters for processing said one of said intermediate
10 frequency band signals to provide a second plurality of intermediate frequency band signals; and
a section of recursive filter resonators for processing said second plurality of intermediate frequency band signals to provide said plurality of further divided frequency band signals.

26. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
15 a midband notch bandpass filter for processing said input digital signals to provide two output signals: a small band signal centered about a frequency midpoint of said input digital signals, and a main wideband signal; and
a bandsplitter for dividing said main wideband signal into two intermediate frequency band signals: a highpass signal and a lowpass signal.

20
27. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
a real bandsplitter for dividing said input digital signals into two intermediate frequency band signals: a real highpass signal and a real lowpass signal; and
a real-to-complex converter for converting said real highpass signal and said real lowpass
25 signal into complex signals.

28. The hearing aid of claim 17, wherein said filter bank analyzer portion comprises:
a real-to-complex converter for converting said input digital signals into complex digital
signals; and
30 a complex bandsplitter for dividing said complex digital signals into two complex intermediate frequency band signals: a complex highpass signal and a complex lowpass signal.

29. The hearing aid of claim 17, wherein said multi-band processor portion
comprises:
35 means for generating a linear instantaneous power estimate for at least one of said individual frequency band signals to produce an instantaneous linear power estimate stream;
means for converting said instantaneous linear power estimate stream into an instantaneous logarithmic power estimate stream;

means for smoothing said instantaneous logarithmic power estimate stream to produce a smoothed logarithmic power estimate stream;

means for calculating a gain coefficient based on said smoothed logarithmic power estimate stream; and

5 means for processing said one of said individual frequency band signals and said gain coefficient to generate one of said plurality of processed frequency band signals.

30. The hearing aid of claim 29, wherein said smoothing means comprises:

10 means for processing said instantaneous logarithmic power estimate stream with a smoothing coefficient time constant to derive a processed power estimate stream;

means for storing a current smoothing filter state;

15 means for processing said current smoothing filter state with said processed power estimate stream to derive a new smoothing filter state, said new smoothing filter state being stored in said storing means, and representing said smoothed logarithmic power estimate stream; and

means for adaptively generating a new smoothing coefficient time constant based on a comparison of said current smoothing filter state with said instantaneous power estimate stream.

31. A hearing aid, comprising:

20 an input transducer for converting audio signals into analog electrical signals;
an analog-to-digital converter for converting said analog signals into real valued input digital signals;

means for converting said real valued input digital signals into complex valued digital signals;

25 means for processing said complex valued digital signals to derive a set of processed complex valued digital signals;

means for converting said processed complex valued digital signals into a set of real valued output digital signals;

30 a digital-to-analog converter for converting said real valued output digital signals into a set of output analog electrical signals; and

an output transducer for converting said output analog electrical signals into a set of output audio signals.

32. The hearing aid of claim 31, wherein said processing means comprises:

35 means for dividing said complex valued digital signals into a plurality of complex individual frequency band signals;

means for processing said plurality of complex individual frequency band signals to derive a plurality of processed complex individual frequency band signals; and

means for combining said plurality of processed complex individual frequency band signals to derive said complex valued digital signals.

33. The hearing aid of claim 32, wherein said means for processing said plurality of complex individual frequency band signals comprises:

means for calculating an instantaneous power estimate for at least one of said plurality of complex individual frequency band signals; and

means for processing said one complex individual frequency band signal according to said instantaneous power estimate.

10

34. A hearing aid, comprising:

an input transducer for converting audio signals into analog electrical signals;

an analog-to-digital converter for converting said analog signals into real valued input digital signals;

15 a bandsplitter for dividing said real valued input digital signals into a plurality of real valued frequency band signals;

a real-to-complex converter for converting at least one of said real valued frequency band signals into a complex frequency band signal;

20 a complex filter bank for further dividing said complex frequency band signal into a plurality of complex frequency sub-band signals;

a multi-band processor for processing said complex frequency sub-band signals to derive a set of processed complex frequency sub-band signals;

25 a complex combiner for combining said processed complex frequency sub-band signals and any of said real valued frequency band signals which were not converted into complex signals to derive a resulting complex signal, said complex combiner further converting said resulting complex signal into a real valued digital output signal;

a digital-to-analog converter for converting said real valued digital output signal into an output analog electrical signal; and

30 an output transducer for converting said output analog electrical signal into an output audio signal.

35. The hearing aid of claim 34, wherein said multi-band processor comprises:

means for calculating an instantaneous power estimate for at least one of said plurality of complex frequency sub-band signals; and

35 means for processing said one complex frequency sub-band signal according to said instantaneous power estimate.

36. A hearing aid, comprising:

an input transducer for converting audio signals into analog electrical signals;

an analog-to-digital converter for converting said analog signals into input digital signals;
a filter bank analyzer for dividing said input digital signals into a plurality of individual frequency band signals, said analyzer comprising:

5 a first section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and
a final section of complementary comb filters for processing said intermediate frequency band signals to provide said plurality of individual frequency band signals;

10 a multi-band processor for processing said plurality of individual frequency band signals to derive a plurality of processed frequency band signals;

a filter bank combiner for combining said plurality of processed frequency band signals to derive said output digital signals.

a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and

15 an output transducer for converting said output analog electrical signals into a set of output audio signals.

37. The hearing aid of claim 36, wherein said multi-band processor comprises:
means for generating a linear instantaneous power estimate for at least one of said
20 individual frequency band signals to produce an instantaneous linear power estimate stream;
means for converting said instantaneous linear power estimate stream into an instantaneous logarithmic power estimate stream;
means for smoothing said instantaneous logarithmic power estimate stream to produce a smoothed logarithmic power estimate stream;
25 means for calculating a gain coefficient based on said smoothed logarithmic power estimate stream; and
means for processing said one of said individual frequency band signals and said gain coefficient to generate one of said plurality of processed frequency band signals.

30 38. The hearing aid of claim 37, wherein said smoothing means comprises:
means for processing said instantaneous logarithmic power estimate stream with a smoothing coefficient time constant to derive a processed power estimate stream;
means for storing a current smoothing filter state;
means for processing said current smoothing filter state with said processed power
35 estimate stream to derive a new smoothing filter state, said new smoothing filter state being stored in said storing means, and representing said smoothed logarithmic power estimate stream; and

means for adaptively generating a new smoothing coefficient time constant based on a comparison of said current smoothing filter state with said instantaneous power estimate stream.

39. A hearing aid, comprising:
an input transducer for converting audio signals into analog electrical signals;
an analog-to-digital converter for converting said analog signals into input digital signals;
5 a filter bank analyzer for dividing said input digital signals into a plurality of individual frequency band signals, said analyzer comprising:
a section of complementary comb filters for processing said input digital signals to provide a plurality of intermediate frequency band signals; and
10 a section of recursive filter resonators for processing said intermediate frequency band signals to provide said plurality of individual frequency band signals;
a multi-band processor for processing said plurality of individual frequency band signals to derive a plurality of processed frequency band signals;
15 a filter bank combiner for combining said plurality of processed frequency band signals to derive a set of output digital signals;
a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and
an output transducer for converting said output analog electrical signals into a set of
20 output audio signals.

40. The hearing aid of claim 39, wherein at least one of said recursive filter resonators comprises a complex one-pole recursive digital filter.

- 25 41. The hearing aid of claim 39, wherein said multi-band processor comprises:
means for generating a linear instantaneous power estimate for at least one of said individual frequency band signals to produce an instantaneous linear power estimate stream;
means for converting said instantaneous linear power estimate stream into an instantaneous logarithmic power estimate stream;
30 means for smoothing said instantaneous logarithmic power estimate stream to produce a smoothed logarithmic power estimate stream;
means for calculating a gain coefficient based on said smoothed logarithmic power estimate stream; and
means for processing said one of said individual frequency band signals and said gain
35 coefficient to generate one of said plurality of processed frequency band signals.

42. The hearing aid of claim 41, wherein said smoothing means comprises:
means for processing said instantaneous logarithmic power estimate stream with a smoothing coefficient time constant to derive a processed power estimate stream;

means for storing a current smoothing filter state;

means for processing said current smoothing filter state with said processed power estimate stream to derive a new smoothing filter state, said new smoothing filter state being stored in said storing means, and representing said smoothed logarithmic power estimate stream;

5 and

means for adaptively generating a new smoothing coefficient time constant based on a comparison of said current smoothing filter state with said instantaneous power estimate stream.

43. A hearing aid, comprising:

10 an input transducer for converting audio signals into analog electrical signals;

an analog-to-digital converter for converting said analog signals into input digital signals;

means for generating a linear instantaneous power estimate for at least a portion of said input digital signals to produce an instantaneous linear power estimate stream;

15 means for converting said instantaneous linear power estimate stream into an instantaneous logarithmic power estimate stream;

means for smoothing said instantaneous logarithmic power estimate stream to produce a smoothed logarithmic power estimate stream;

means for receiving and processing said smoothed logarithmic power estimate stream and said input digital signals to derive a set of output digital signals;

20 a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and

an output transducer for converting said output analog electrical signals into a set of output audio signals.

25 44. The hearing aid of claim 43, wherein said smoothing means comprises:

means for processing said instantaneous logarithmic power estimate stream with a smoothing coefficient time constant to derive a processed power estimate stream;

means for storing a current smoothing filter state;

30 means for processing said current smoothing filter state with said processed power estimate stream to derive a new smoothing filter state, said new smoothing filter state being stored in said storing means, and representing said smoothed logarithmic power estimate stream; and

means for adaptively generating a new smoothing coefficient time constant based on a comparison of said current smoothing filter state with said instantaneous power estimate stream.

35

45. A hearing aid, comprising:

an input transducer for converting audio signals into analog electrical signals;

an analog-to-digital converter for converting said analog signals into input digital signals;

a comb filter section for processing said input digital signals to partition said input digital signals across a frequency spectrum into a plurality of non-overlapping bandpass spectral lobes;

a lobe selection filter section comprising a plurality of separate bandpass filters, each of said bandpass filters being tuned in frequency to select and process one of said bandpass spectral lobes, said lobe selection filter section outputting a plurality of separate bandpass signals;

a bandpass combiner for combining said plurality of separate bandpass signals to derive a set of output digital signals;

a digital-to-analog converter for converting said output digital signals into a set of output analog electrical signals; and

an output transducer for converting said output analog electrical signals into a set of output audio signals.

46. The hearing aid of claim 45, wherein said lobe selection filter section comprises a recursive filter resonator for processing one of said bandpass spectral lobes.

47. The hearing aid of claim 46, wherein said recursive filter resonator comprises a complex one-pole recursive digital filter.

48. A digital signal processing hearing aid, comprising:
an input transducer for converting audio signals into analog electrical signals;
an analog-to-digital converter for converting said analog electrical signals into a digital input stream of numerical values;
a storage for storing said numerical values;
an arithmetic logic unit coupled to said storage for performing digital signal processing operations on said numerical values to filter said digital input stream, said arithmetic logic unit comprising:

an adder unit having an input for receiving said numerical values, and an output;
a multiple input multiplier unit having at least one input coupled to the output of said adder unit, and an output; and
an accumulation unit having an input coupled to the output of said multiplier unit, and an output for providing a digital output stream;
a digital-to-analog converter coupled to said arithmetic logic unit for converting said digital output stream into a set of output analog electrical signals; and
an output transducer for converting said output analog electrical signals into a set of output audio signals.

49. A hearing aid, comprising:
an input transducer for converting audio signals into analog electrical signals;

an analog-to-digital converter for converting said analog electrical signals into a digital input stream of numerical values;

multiple stream producing means for processing said stream of numerical values to produce at least one additional stream of numerical values;

5 numerical stream smoothing means for filtering at least one of said additional streams of numerical values to produce smoothed streams of numerical values, wherein a time span over which the smoothing occurs is determined by at least one numerical time constant;

time constant adaptation means for changing the value of said numerical time constant based on values derived from said additional streams of numerical values and said smoothed
10 streams of numerical values;

output stream producing means for producing an output stream of numerical values based on said additional streams of numerical values and said smoothed streams of numerical values;

a digital-to-analog converter for converting said output stream of numerical values into a set of output analog electrical signals; and

15 an output transducer for converting said output analog electrical signals into a set of output audio signals.

50. A digital signal processing hearing aid, comprising:

an input transducer for converting audio signals into analog electrical signals;

20 an analog to digital converter for converting said analog electrical signals into a real valued digital input stream of numerical values;

filter bank means for generating a plurality of bandpass signals from said input stream of numerical values, said filter bank means comprising means for dynamically modifying the number and bandwidth of said bandpass signals based on numerical control variables associated
25 with said filter bank means;

bandpass processing means for processing and combining said plurality of bandpass signals to produce a real-valued digital output stream of numerical values;

a digital-to-analog converter for converting said real-valued digital output stream into a set of output analog electrical signals; and

30 an output transducer for converting said output analog electrical signals into a set of output audio signals.

1/22

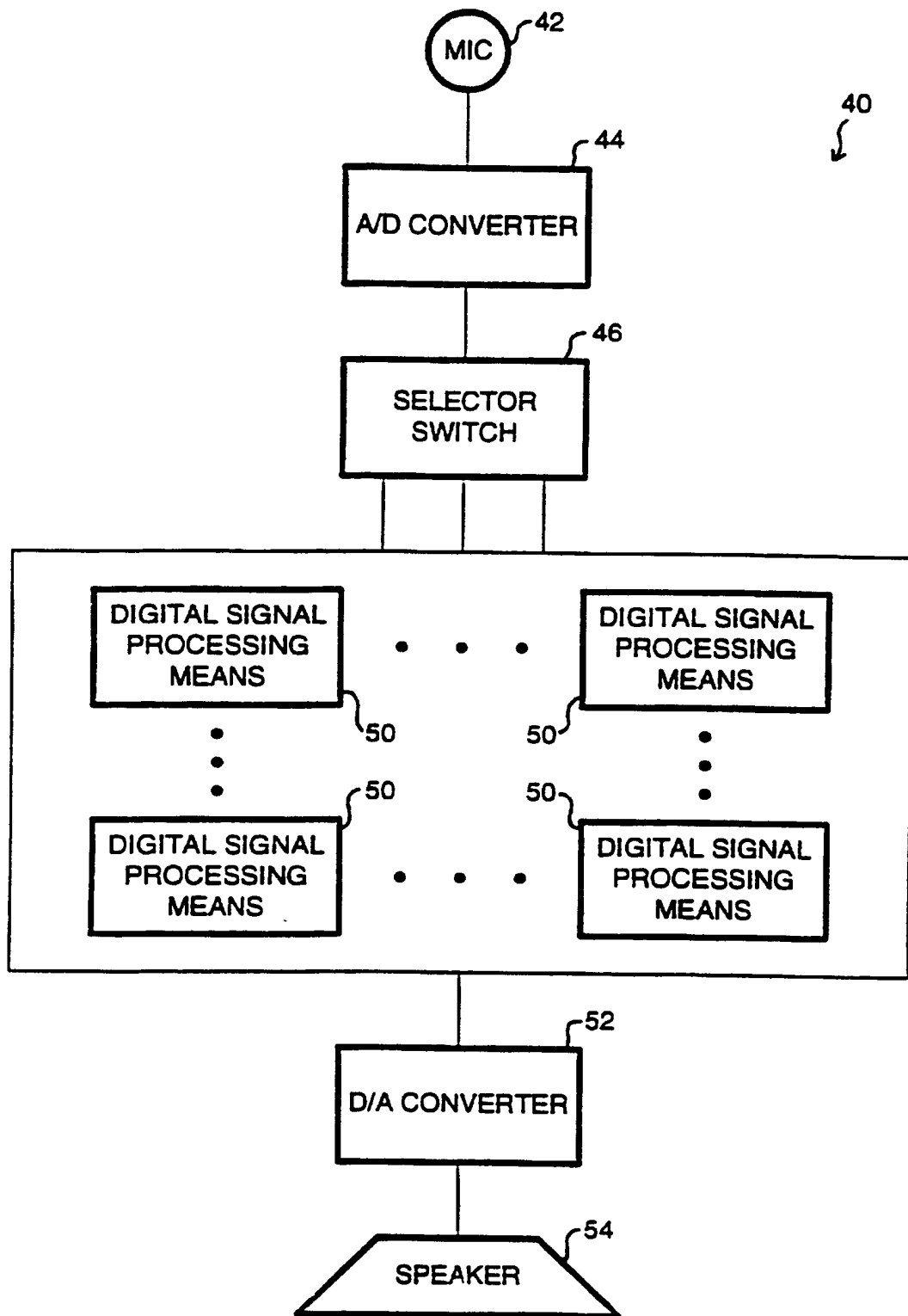


Figure 1a

2/22

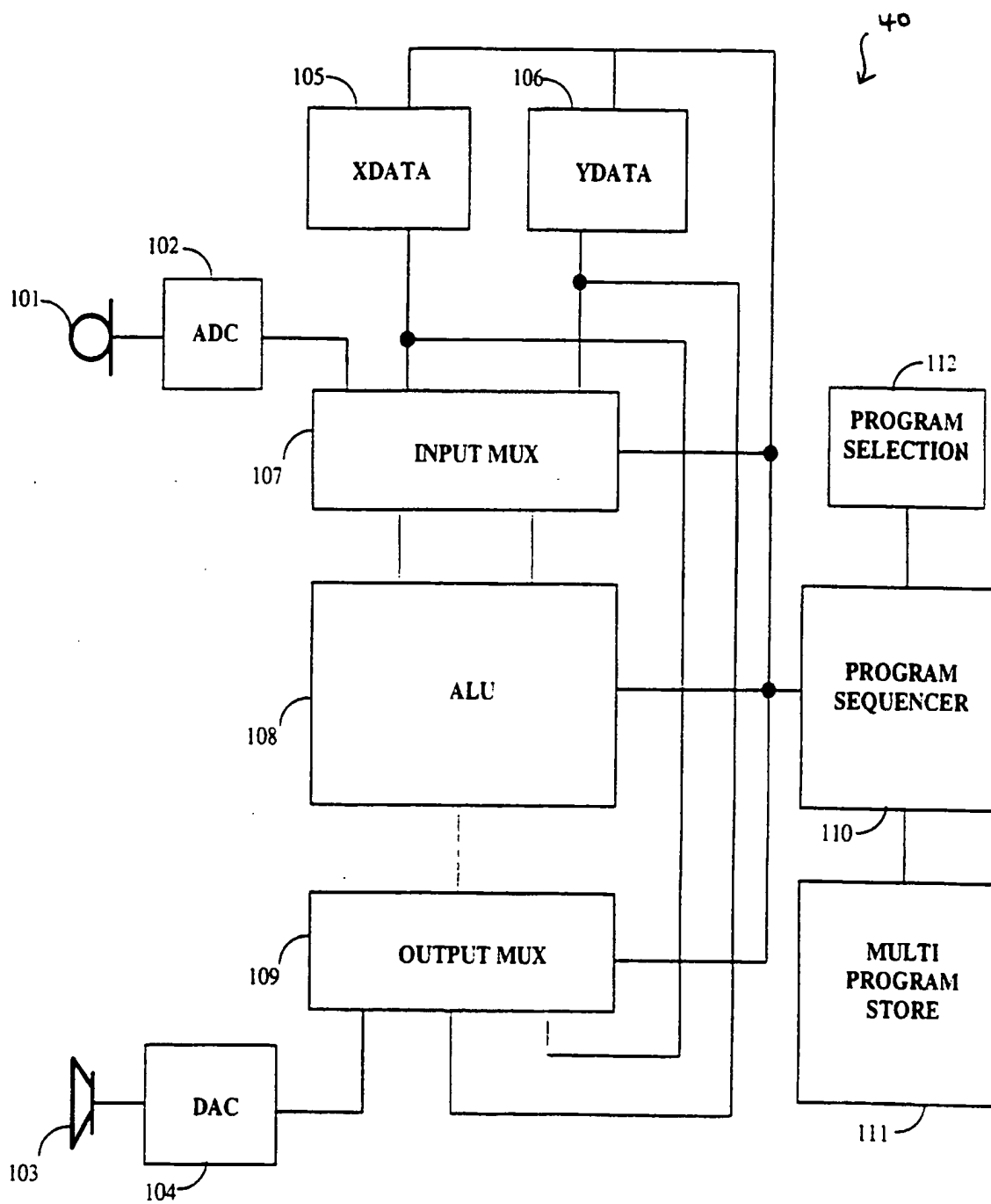


Figure 1 b

3/22

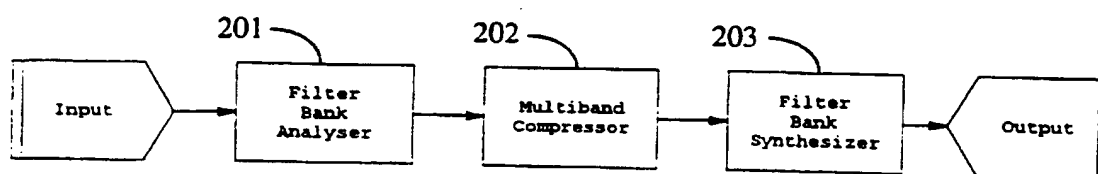


Figure 2

4/22

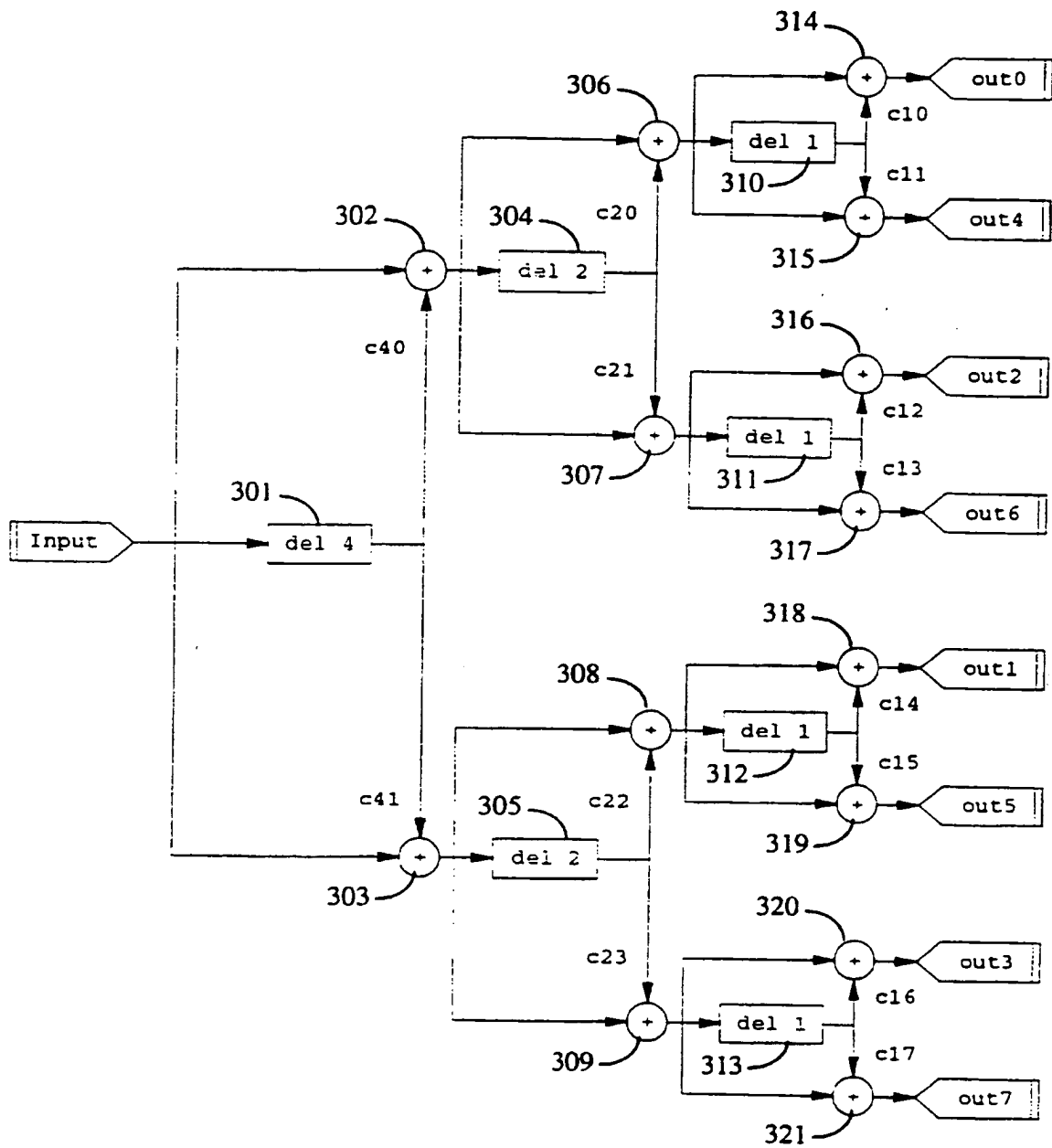


Figure 3

5/22

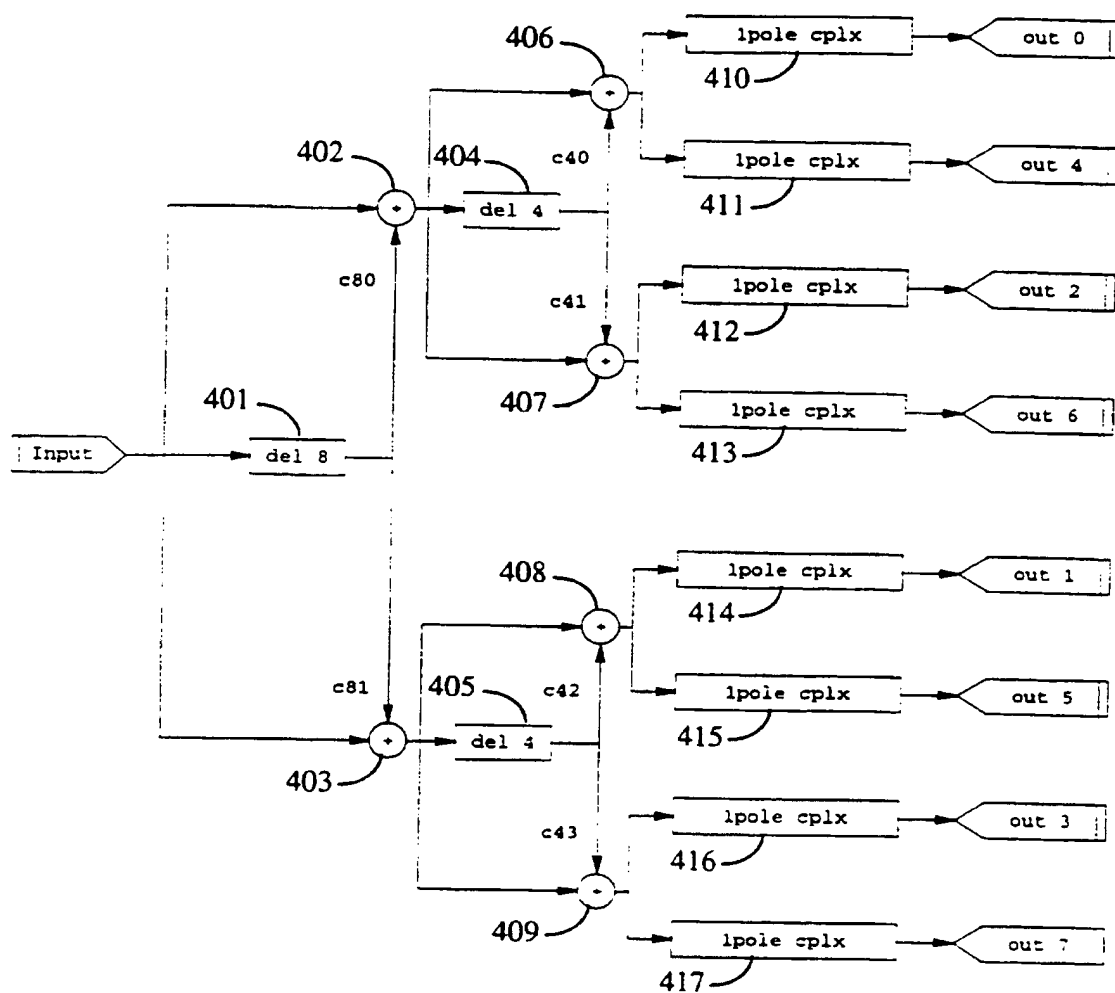


Figure 4

6/22

Figure 5

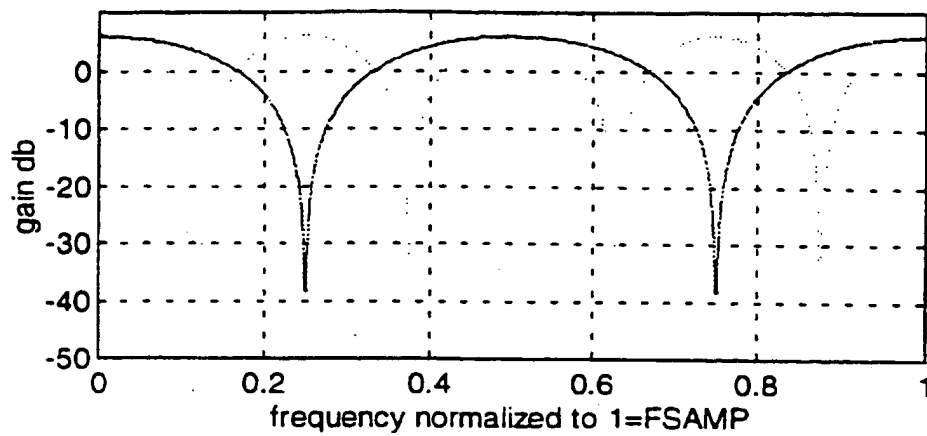
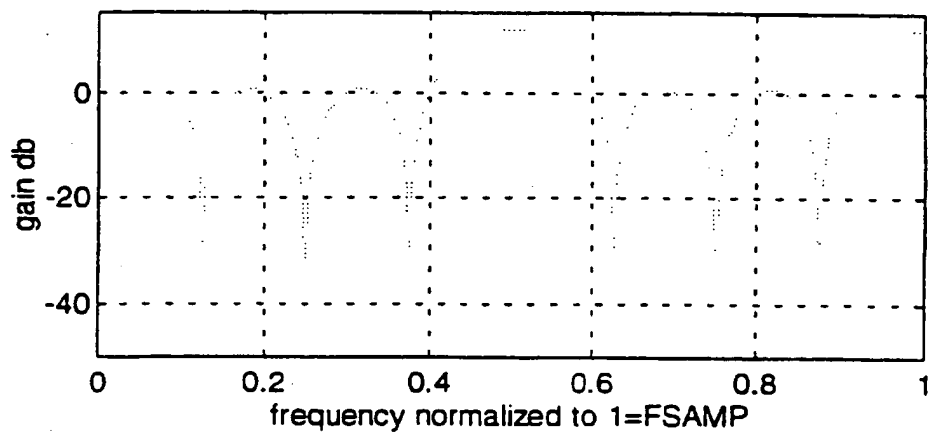


Figure 6



7/22

Figure 7

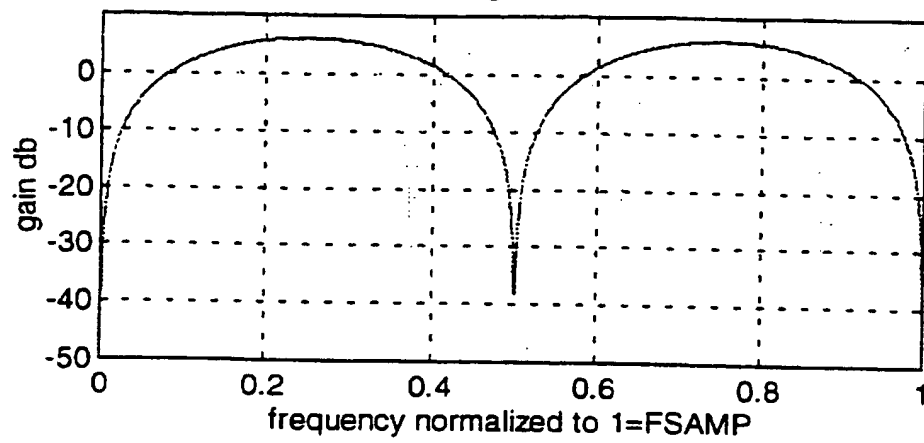
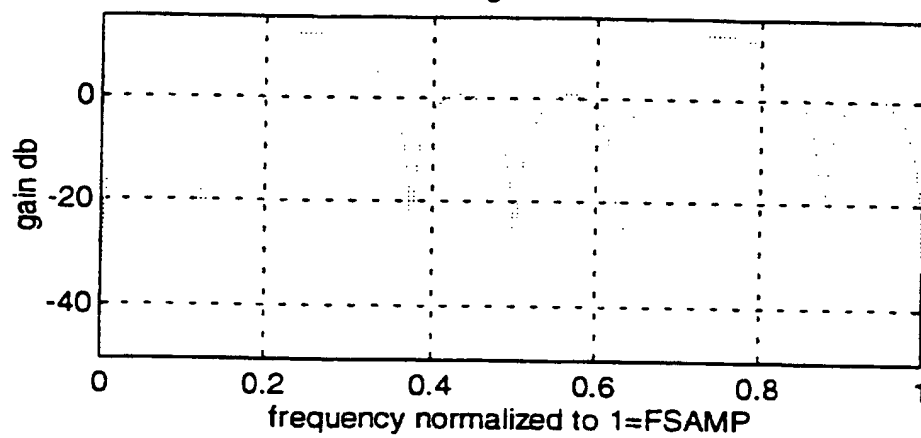


Figure 8



8/22

Figure 9

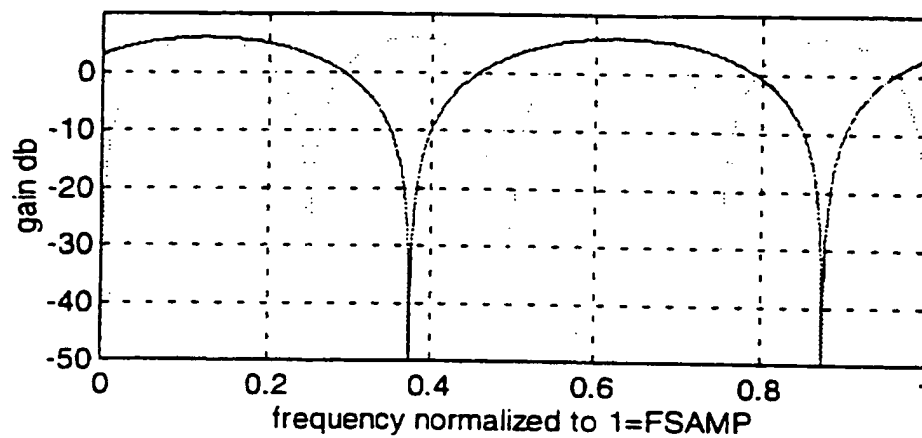
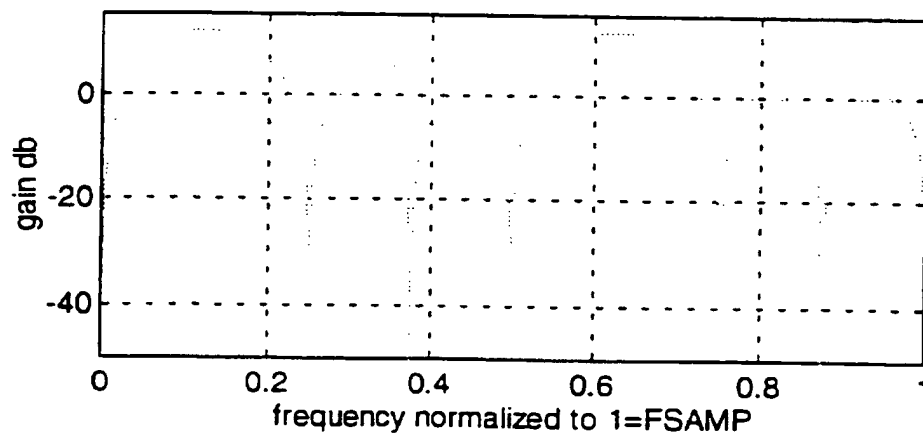


Figure 10



9/22

Figure 11

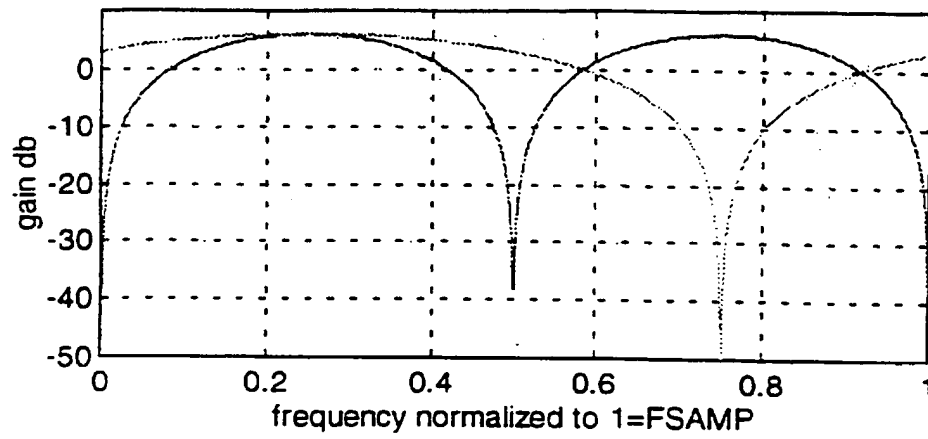
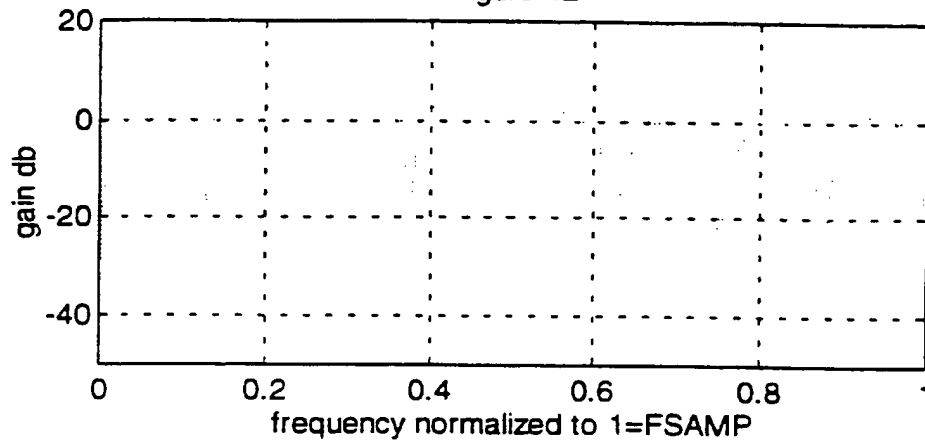


Figure 12



10/22

Figure 13

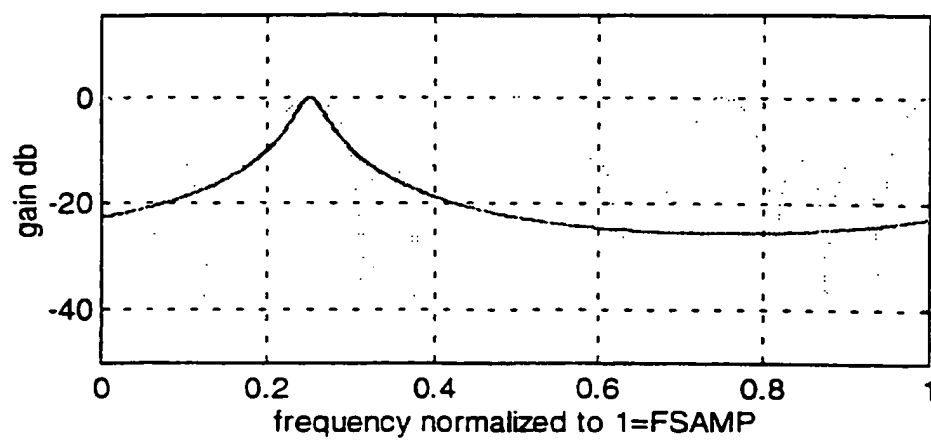
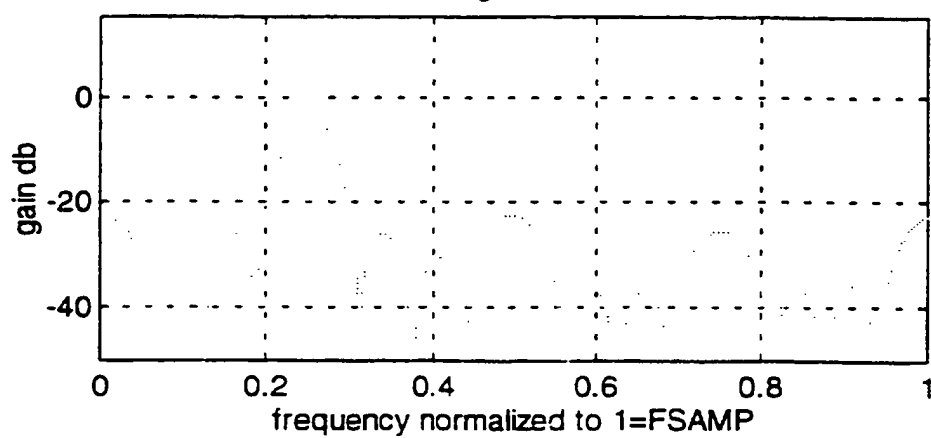
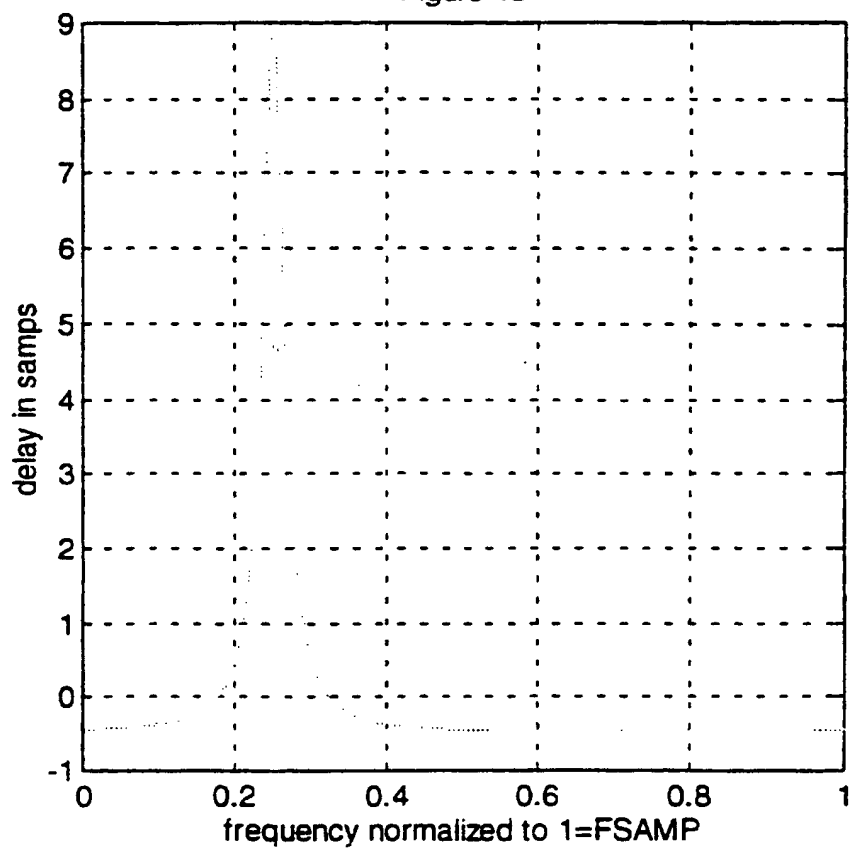


Figure 14



11/22

Figure 15



12/22

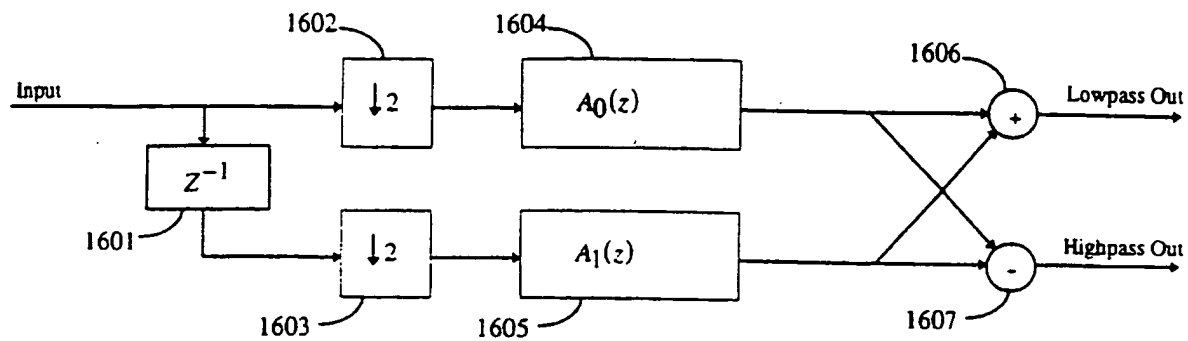


Figure 16

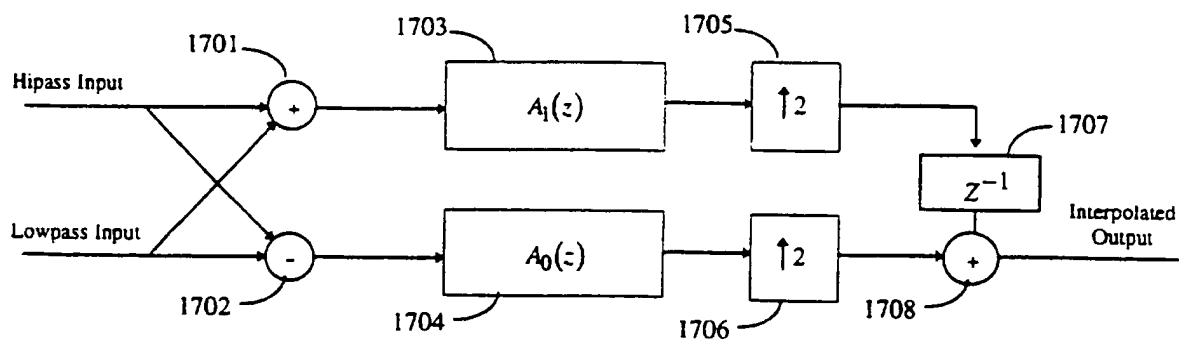


Figure 17

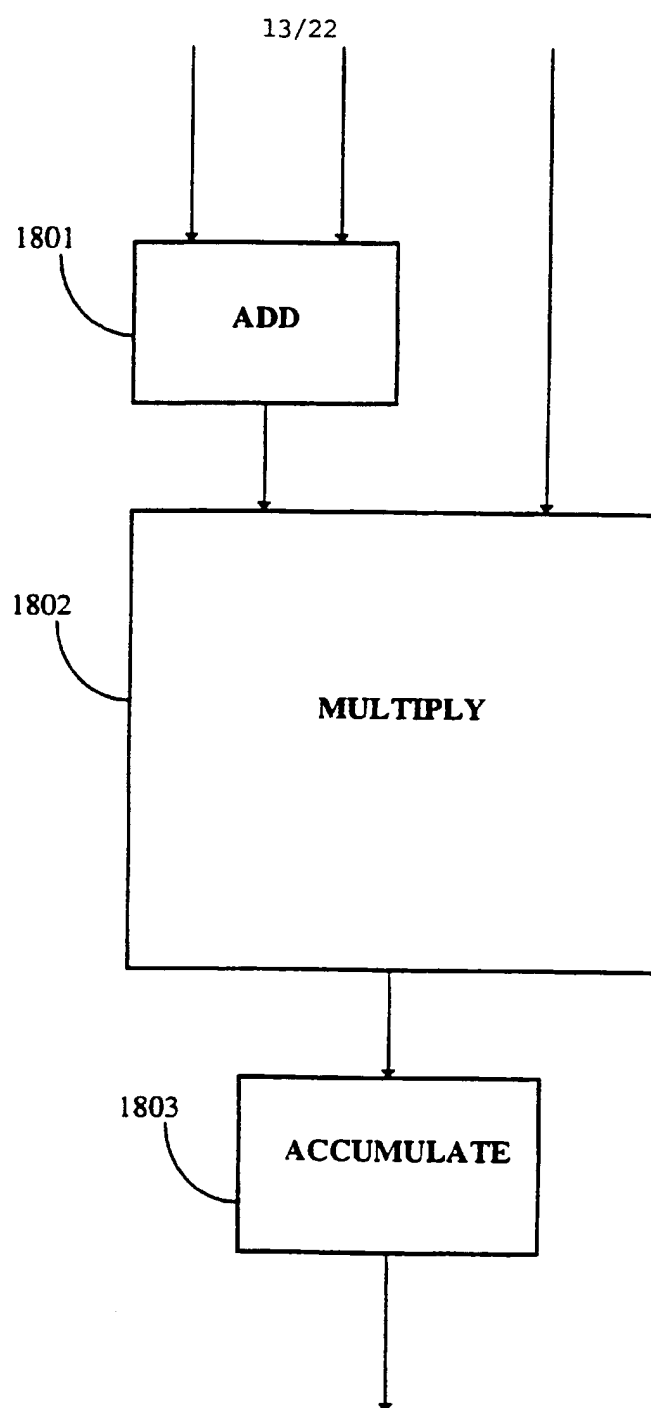


Figure 18

14/22

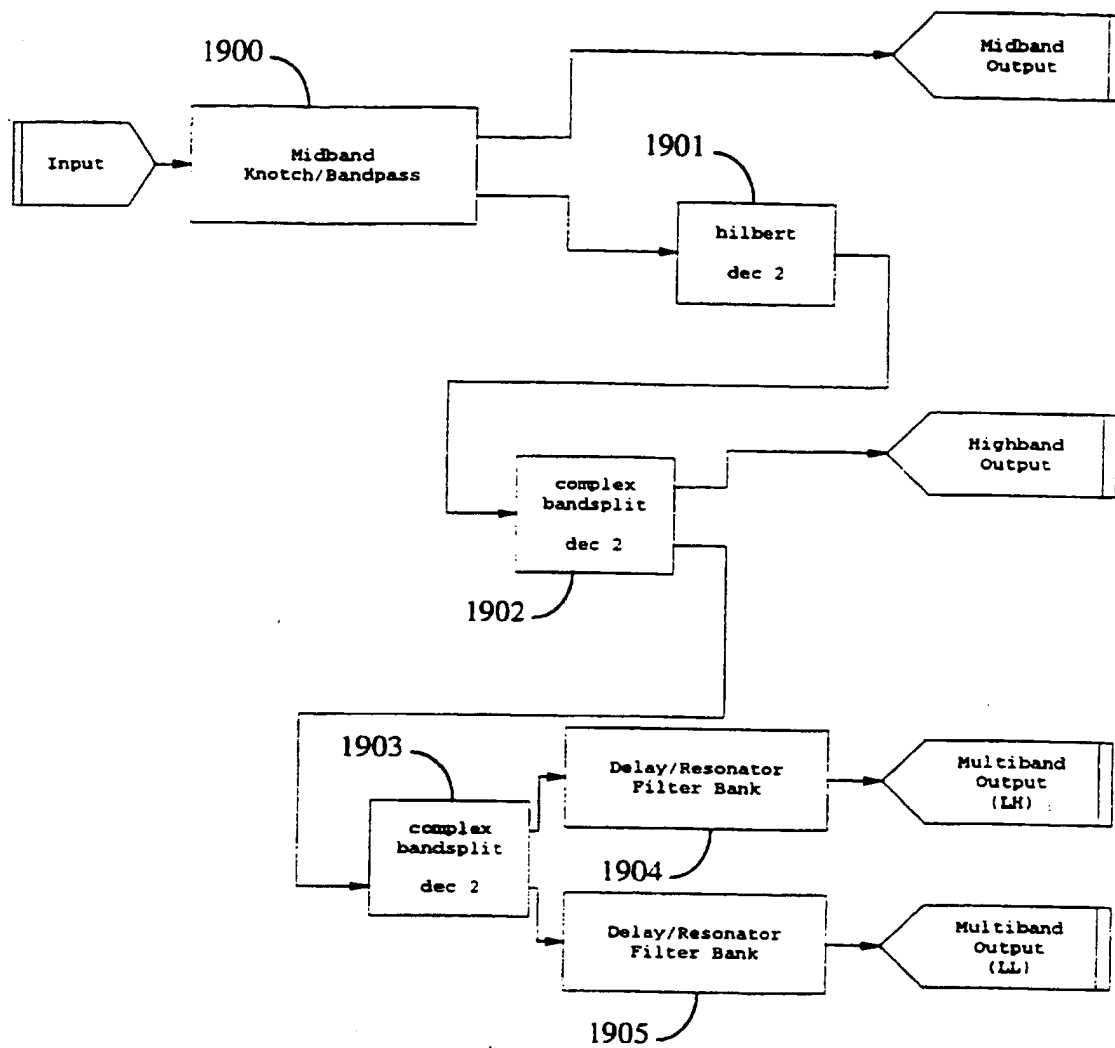


Figure 19

15/22

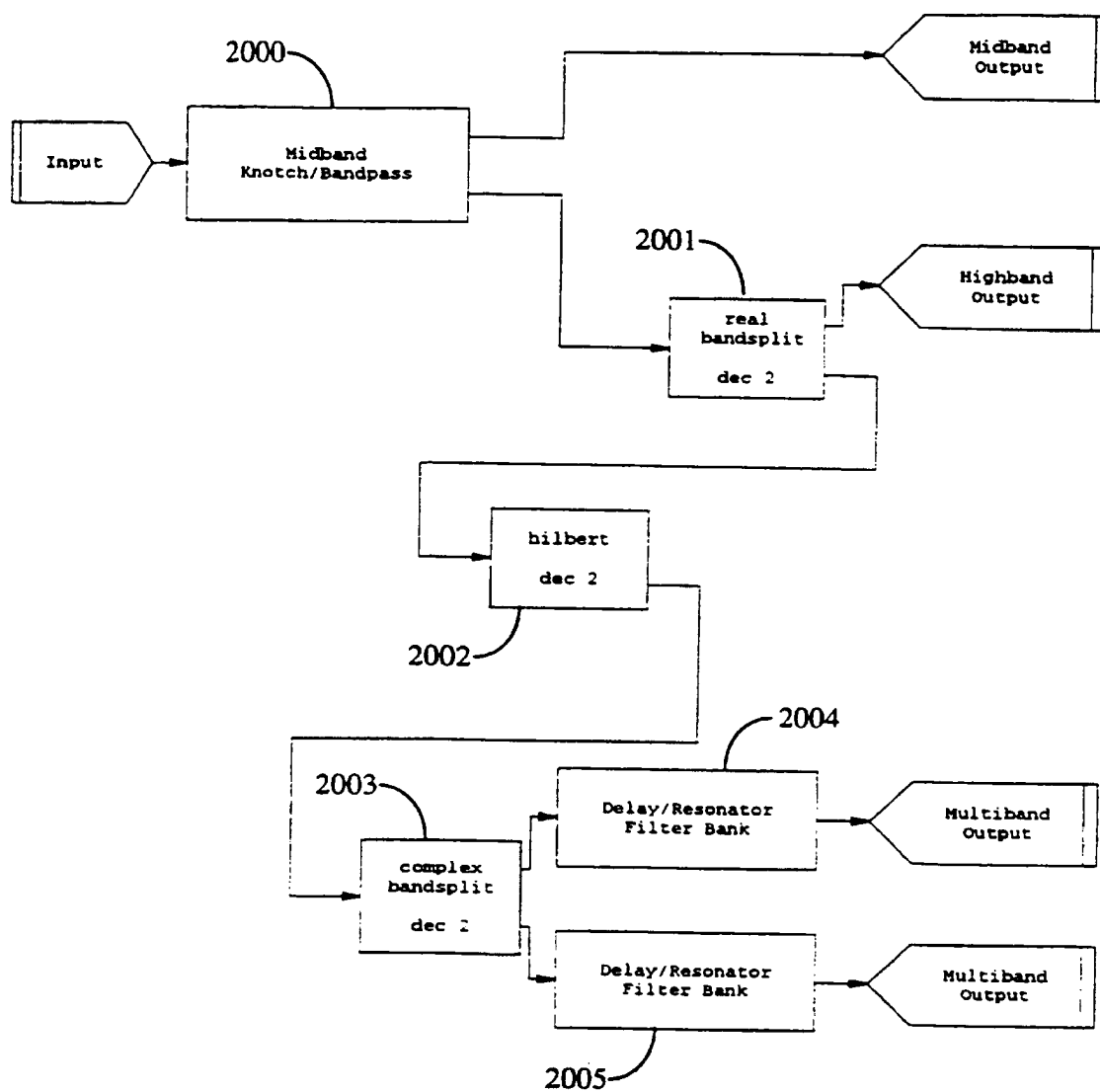


Figure 20

16/22

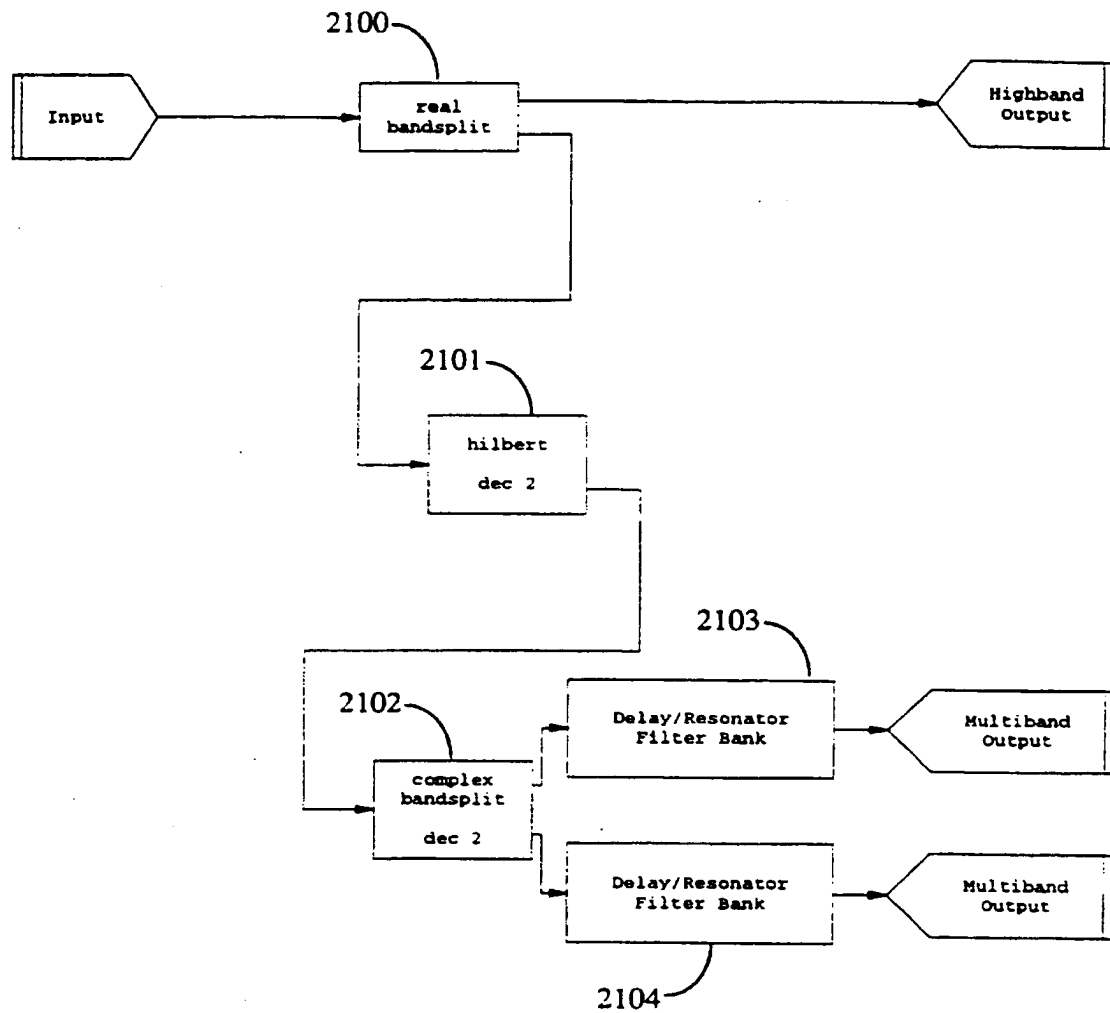


Figure 21

17/22

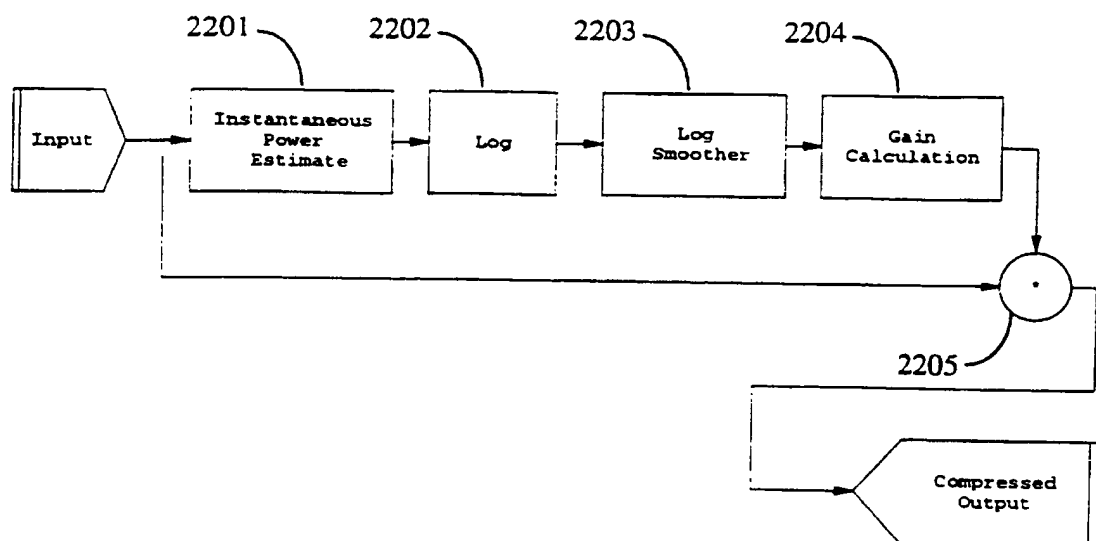


Figure 22

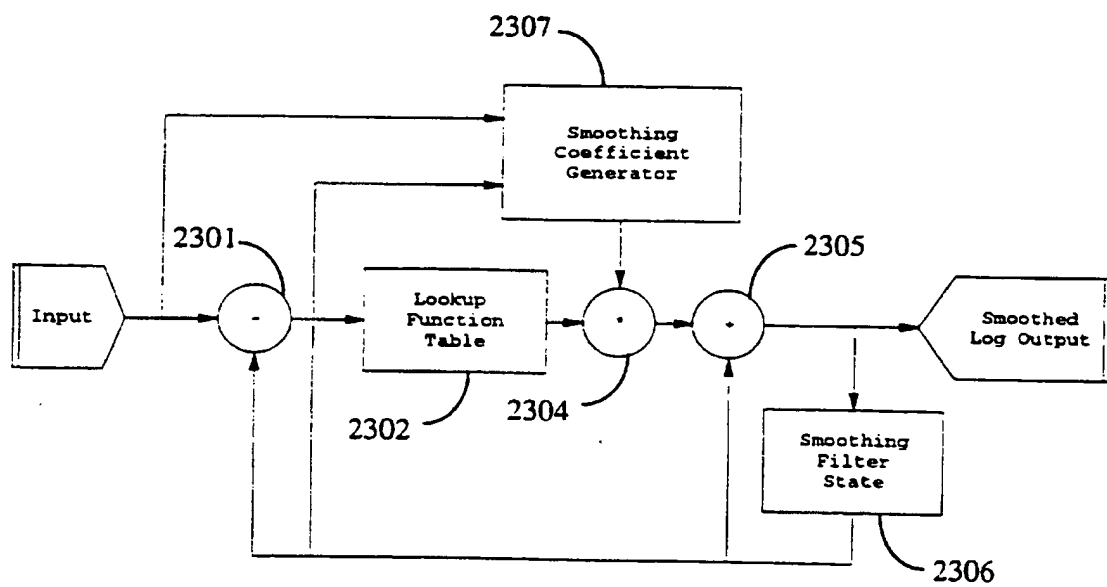


Figure 23

18/22

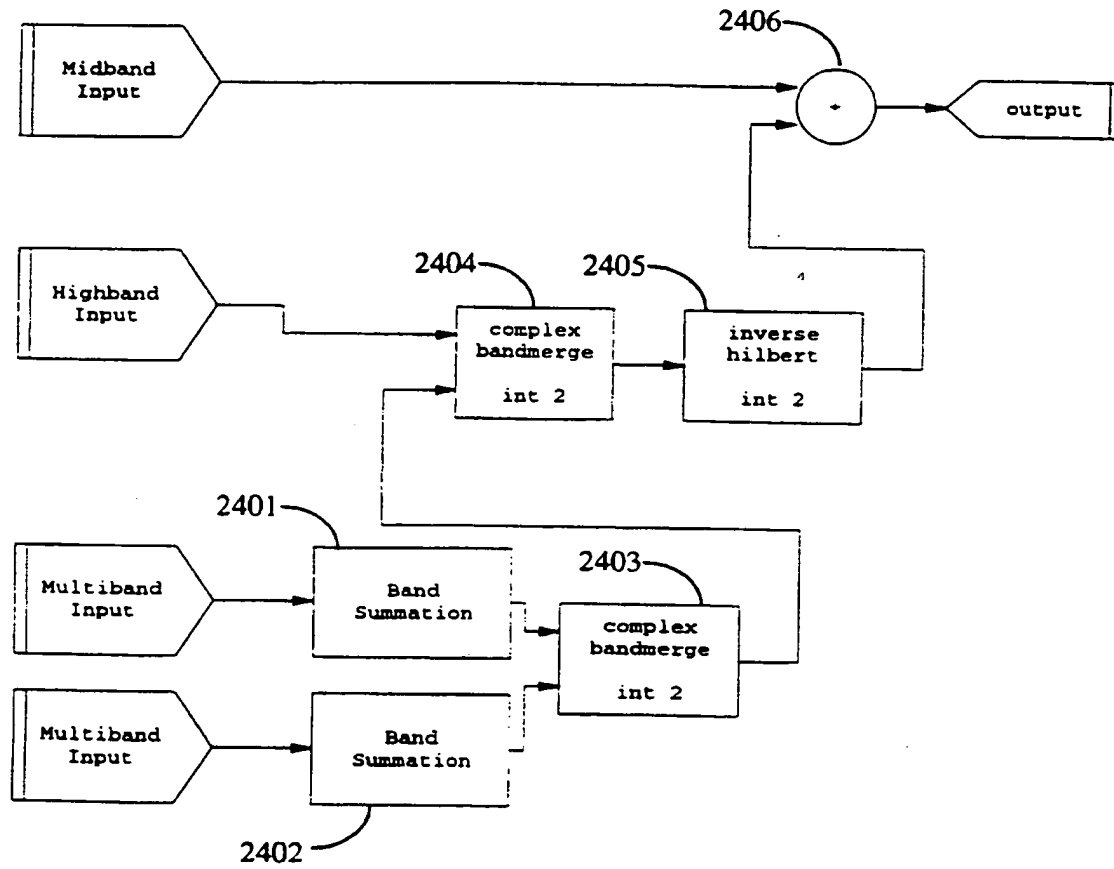


Figure 24

19/22

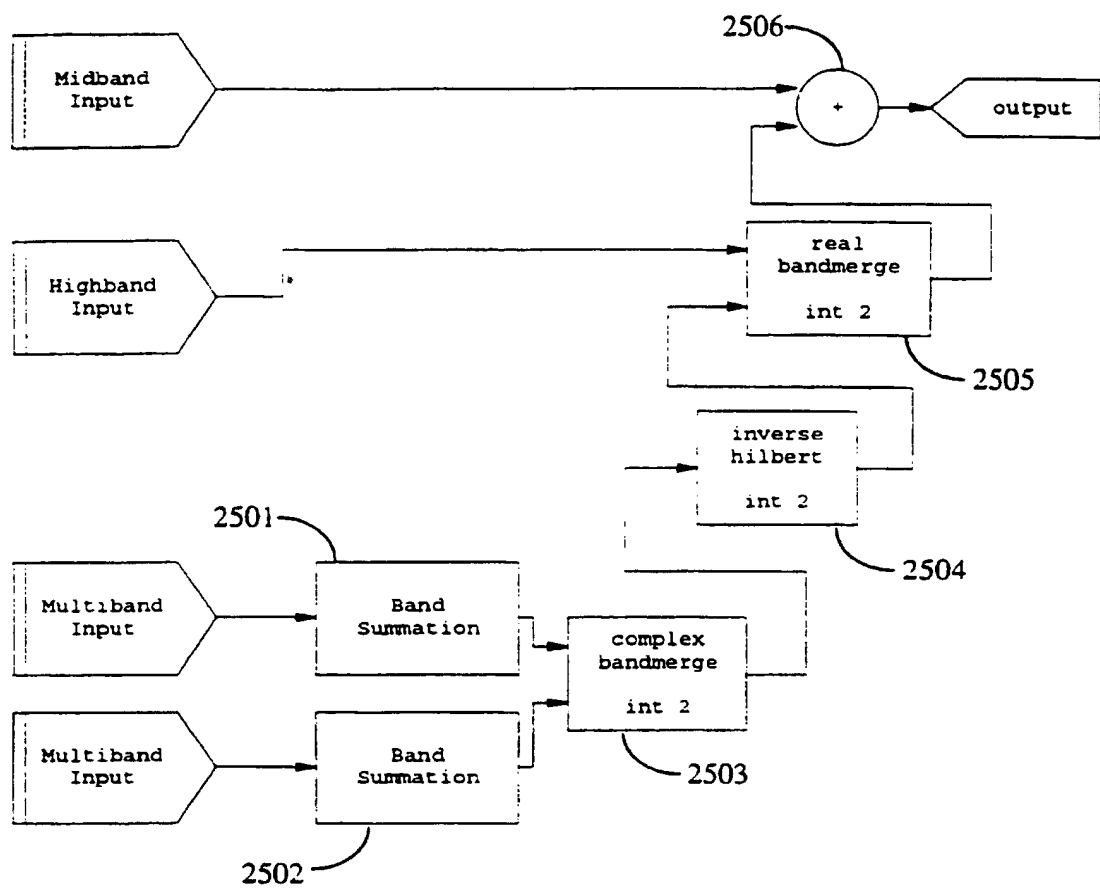


Figure 25

20/22

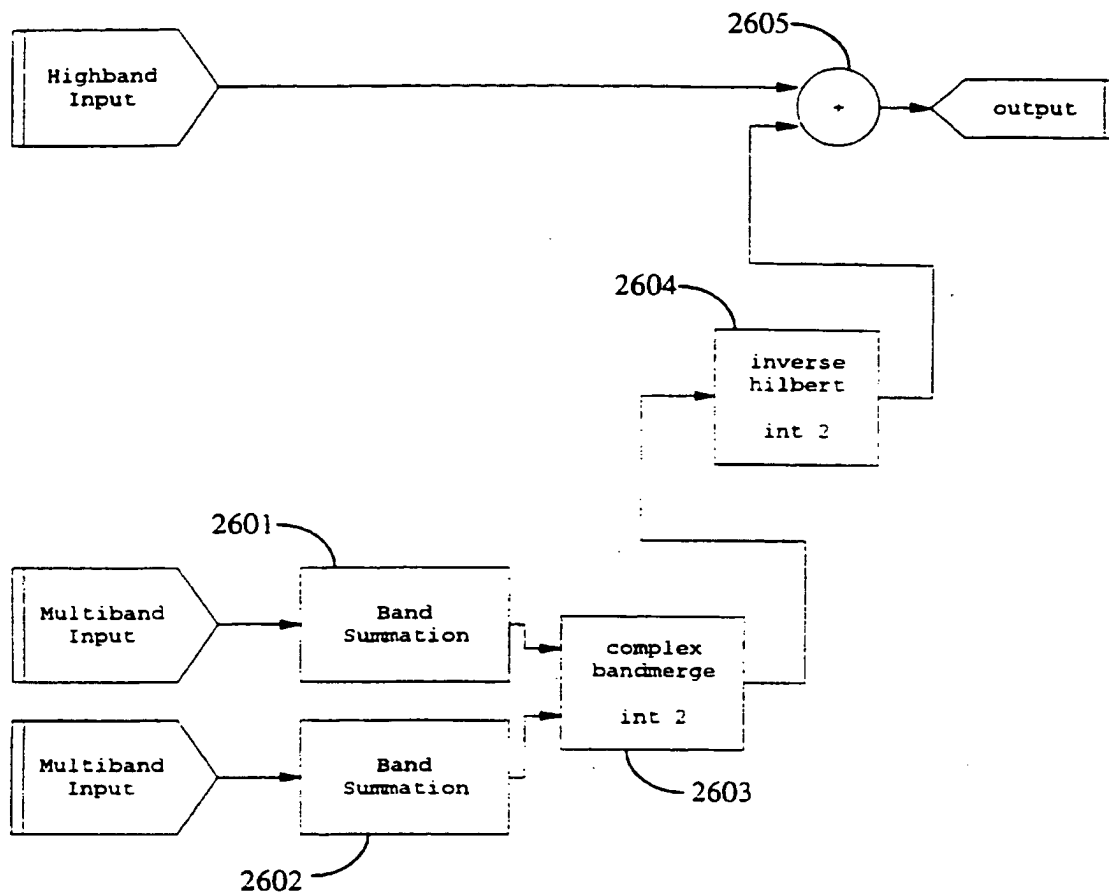


Figure 26

Figure 27

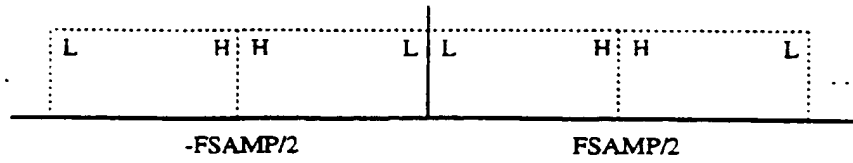


Figure 28

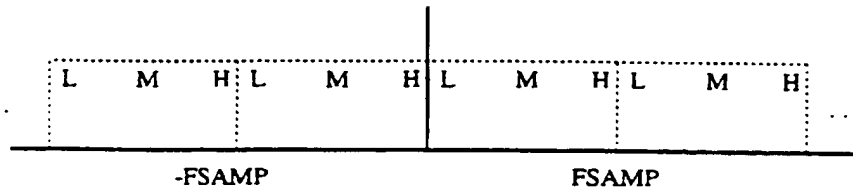


Figure 29

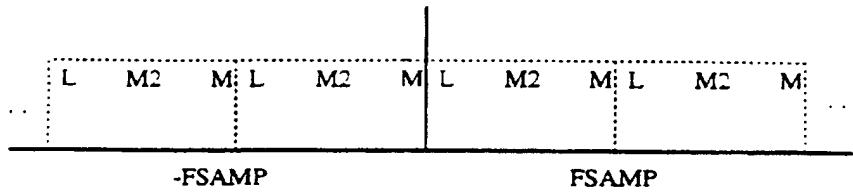


Figure 30

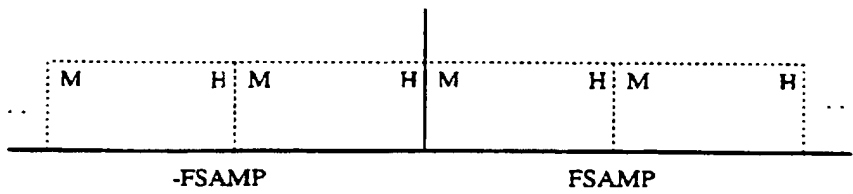


Figure 31

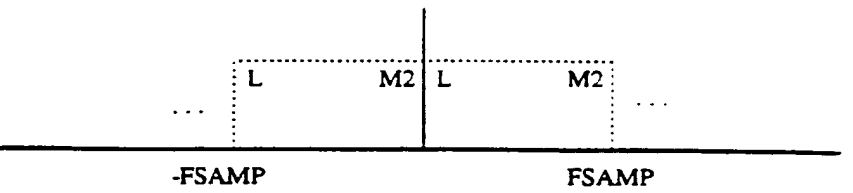
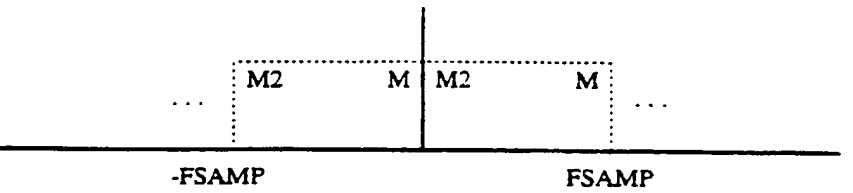


Figure 32



22/22

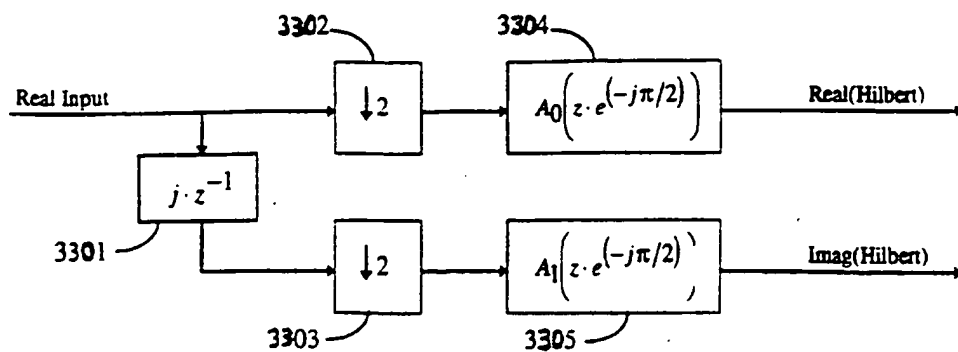


Figure 33

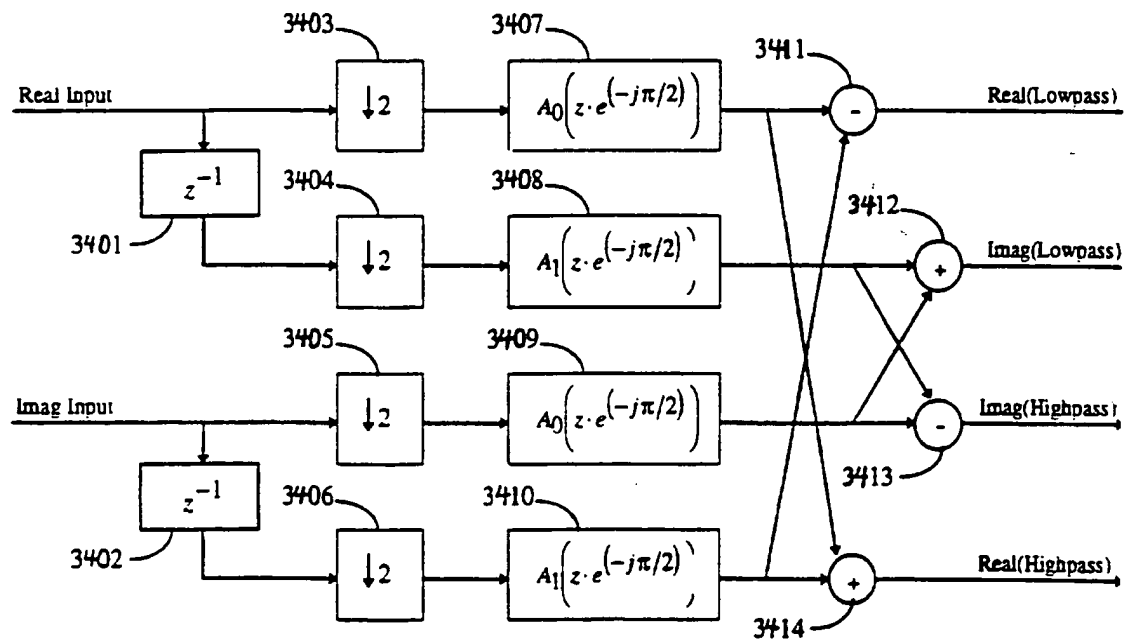


Figure 34

THIS PAGE BLANK (USPTO)